Biomechanics of sprint running : a methodological contribution

Elena Bergamini

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Administrative Universities: Università degli Studi di Bologna (Bologna) – Arts et Métiers ParisTech (Paris)

Associate University: Università degli Studi di Roma "Foro Italico", ex Istituto Universitario di Scienze Motorie (Roma)

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Elena Bergamini

Supevisors: **Prof. Aurelio Cappozzo**
Università degli Studi di Roma "Foro Italico"

Prof. Wafa Skalli
Arts et Métiers ParisTech

Dr. Valentina Camomilla
Università degli Studi di Roma "Foro Italico"

Dr. Hélène Pillet
Arts et Métiers ParisTech

Coordinator: **Prof. Angelo Cappello**
Università degli Studi di Bologna

Co-examiner: **Prof. Luca Cristofolini**
Università degli Studi di Bologna

Alma Mater Studiorum – Università di Bologna
Università degli Studi di Roma "Foro Italico"
Arts et Métiers ParisTech

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BIOMECHANICS OF SPRINT RUNNING: A METHODOLOGICAL CONTRIBUTION

Presentata da: **Elena Bergamini**

Coordinatore Dottorato

Prof. Angelo Cappello

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Relatori

Prof. Aurelio Cappozzo

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Prof.ssa Wafa Skalli

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PREFACE

Sports biomechanics uses the scientific methods of mechanics to study the internal and external forces acting on a human body and the effects produced by these forces in sports activities. It is, therefore, concerned, with the ways in which sports movements are performed, often referred to as sports techniques.

With this respect, sports biomechanics has an applicative nature and it has an ultimate objective: the achievement of an effective technique. Indeed, while it is often believed that the main goal of sports biomechanists and coaches is to obtain the best athlete's performance, it must be kept in mind that performance enhancement and injury risk are strictly related. If improving the athlete's performance would entail increasing the risk of injury, no actual effective improvement would be obtained. This is the reason why sports biomechanics is often described as having two aims that may be perceived as incompatible: the reduction of the risk of injury (such as identification of the causes of back injuries in cricket, or the causes of knee joint injuries in sprint running) and the improvement of performance (such as optimising gymnastics performance through simulation of in-flight movements, or studying the effects of tennis racquet stiffness on the performance of young athletes).

Although sports biomechanists and coaches, as well as athletes and physicians, share exactly the same objectives, there is still a gap between researchers and practitioners. Coaches commonly base their evaluation of athletes' performance and of the efficacy of their training program on competition results, field tests and qualitative visual inspection. Such evaluations are often readable, easy to use and they provide information about the global performance. Nevertheless, they are often not able to establish an understanding of causal mechanisms for the selected movements (such as the role of internal rotation of the upper arm in hitting or striking, or the influence of elastic energy and muscle pre-stretch in stretch-shorten-cycle actions). On the other hand, biomechanists, accused to be over-concerned with methodology, often lack of pragmatism.

In order to bridge the gap between the two worlds, the establishment of a common and understandable language is crucial. The importance of quantitative assessment of athletes' motor skills needs to be well perceived by coaches, as well as by athletes and sports physicians. At the same time, biomechanists must be able to fully understand the actual needs of practitioners and find the appropriate way to communicate and propose their results.

There are two main limitations that have to be overcome in order to achieve these goals: first, the difficulties in performing in-field athlete evaluation and in obtaining data in a natural setting such as competition, without influencing or constraining athletes' activities. To date, in fact, sports biomechanical analysis is typically performed by means of stereophotogrammetry, force platforms, high-speed cameras, or optical contact time meters. Such solutions, however, are expensive, characterized by a cumbersome and time-consuming experimental set up and tend to constrain the subject and the analysed motor task. Second, there is a lack of methods specifically designed for sports applications, even when dealing with marker-based motion analysis. Protocols commonly developed for clinical biomechanical assessment, in fact, could hardly be transposed to the analysis of sports motor acts, which are characterized by higher accelerations and explosiveness with respect to the former.

In this framework, the present thesis aims at providing a contribution in these directions, focusing on the development of methodologies which could help in overcoming the above mentioned limitations, filling the gap between researchers and practitioners.

ABSTRACT

ENGLISH VERSION

Sports biomechanics describes human movement from a performance enhancement and an injury reduction perspective. In this respect, the purpose of sports scientists is to support coaches and physicians with reliable information about athletes' technique. The lack of methods allowing for in-field athlete evaluation as well as for accurate joint force estimates represents, to date, the main limitation to this purpose. The investigations illustrated in the present thesis aimed at providing a contribution towards the development of the above mentioned methods.

Two complementary approaches were adopted: a Low Resolution Approach – related to performance assessment – where the use of wearable inertial measurement units is exploited during different phases of sprint running, and a High Resolution Approach – related to joint kinetics estimate for injury prevention – where subject-specific, non-rigid constraints for knee joint kinematic modelling used in multi-body optimization techniques are defined.

Results obtained using the Low Resolution Approach indicated that, due to their portability and inexpensiveness, inertial measurement systems are a valid alternative to laboratory-based instrumentation for in-field performance evaluation of sprint running. Using acceleration and angular velocity data, the following quantities were estimated: trunk inclination and angular velocity, instantaneous horizontal velocity and displacement of a point approximating the centre of mass, and stride and support phase durations.

As concerns the High Resolution Approach, results indicated that the length of the anterior cruciate and lateral collateral ligaments decreased, while that of the deep bundle of the medial collateral ligament increased significantly during flexion. Variations of the posterior cruciate and the superficial bundle of the medial collateral ligament lengths were concealed by the experimental indeterminacy. A mathematical model was provided that allowed the estimate of subject-specific ligament lengths as a function of knee flexion and that can be integrated in a multi-body optimization procedure.

ITALIAN VERSION

La biomeccanica dello sport descrive il movimento umano con l'obiettivo di migliorare la prestazione atletica e di ridurre l'incidenza di infortuni. In questo contesto, lo scopo degli esperti di scienze dello sport è quello di fornire ad allenatori e medici informazioni affidabili sulla tecnica di esecuzione del gesto sportivo in esame. La mancanza di metodi che consentano la valutazione dell'atleta direttamente sul campo e la stima accurata della dinamica articolare costituisce, ad oggi, il principale limite nel raggiungimento di questo scopo. La presente tesi si propone di fornire un contributo allo sviluppo di tali metodi.

Il lavoro si articola secondo due approcci complementari: un Approccio a Bassa Risoluzione - legato alla valutazione della prestazione – attraverso il quale è stato esplorato l'uso di sensori inerziali indossabili durante diverse fasi della corsa di velocità, e un Approccio ad Alta Risoluzione - relativo alla stima della dinamica articolare per la prevenzione degli infortuni - dove sono stati definiti vincoli non rigidi per un modello cinematico del ginocchio da integrare in tecniche di ottimizzazione multi-segmento per la stima della posizione e dell'orientamento delle ossa durante il movimento.

I risultati ottenuti con l'Approccio a Bassa Risoluzione indicano che, in virtù della loro portabilità ed economicità, i sensori inerziali rappresentano una valida alternativa alla tradizionale strumentazione di laboratorio per la valutazione della prestazione durante la corsa. Utilizzando i dati di accelerazione e velocità angolare provenienti dai sensori, sono stati stimati l'inclinazione e la velocità angolare del tronco, la velocità lineare istantanea e lo spostamento di un punto che approssima il centro di massa, e le durate della fase di appoggio e del ciclo del passo.

Per quanto riguarda l'Approccio ad Alta Risoluzione, i risultati indicano che le lunghezze del legamento crociato anteriore e del collaterale laterale diminuiscono, mentre quella del fascio profondo del legamento collaterale mediale aumenta durante la flessione. Le variazioni di lunghezza del legamento crociato posteriore e del fascio superficiale del legamento collaterale mediale sono risultate dello stesso ordine dell'errore sperimentale. Al fine di integrare tali informazioni in una procedura di ottimizzazione multi-segmento, è stato definito un modello matematico del ginocchio che fornisce le lunghezze plausibili dei legamenti in funzione dell'angolo di flessione.

FRENCH VERSION

La biomécanique du sport décrit le mouvement humain dans le but d’améliorer la performance et de réduire les blessures. Dans ce contexte, le but des experts des sciences sportives est de fournir aux entraîneurs et médecins des informations fiables sur la technique des athlètes. Le manque de méthodes permettant l'évaluation des athlètes sur le terrain ainsi que l'estimation précise des efforts articulaires représente, à ce jour, une limitation majeure pour atteindre ces objectifs. Les travaux effectués dans la thèse vise à contribuer au développement des ces méthodes.

Deux approches complémentaires ont été adoptées: une Approche à Basse Résolution – relative à l'évaluation de la performance – où l'utilisation de capteurs inertiels portables est exploitée au cours des différentes phases de la course de vitesse, et une Approche à Haute Résolution – lié à l’estimation des efforts articulaires pour la prévention des blessures – où des contraintes personnalisées pour la modélisation cinématique du genou dans le contexte des techniques d'optimisation multi-corps ont été définies.

Les résultats obtenus par l'Approche à Basse Résolution indiquent que, en raison de leur portabilité et leur faible coût, les capteurs inertiels sont une alternative valable aux instrumentations de laboratoire pour l'évaluation de la performance pendant la course de vitesse. En utilisant les données d'accélération et de vitesse angulaire, l’inclinaison et la vitesse angulaire du tronc, la vitesse horizontale instantanée et le déplacement du centre de masse, ainsi que la durée de la phase d’appui et du pas ont été estimés.

En ce qui concerne l'Approche à Haute Résolution, les résultats ont montré que les longueurs du ligament antérieur croisé et du latéral externe diminuaient, alors que celle du faisceau profond du ligament latéral interne augmentait de manière significative lors de la flexion. Les variations de longueur du ligament croisé postérieur et du faisceau superficiel du ligament latéral médial étaient de l’ordre de l'indétermination expérimentale. Un modèle mathématique a été fourni qui a permis l'estimation des longueurs ligamentaires personnalisées en fonction de la flexion du genou et qui peuvent être intégrées dans une procédure d'optimisation multi-corps.

EXTENDED SUMMARY

INTRODUCTION AND AIM OF THE THESIS

Sports biomechanics describes the human movement from a performance enhancement and an injury reduction perspective. In this respect, the purpose of sports scientists is to support coaches and physicians with reliable and usable information related to the athletes' correct or incorrect technique.

Biomechanical research in sports has usually produced interesting descriptions of the basic kinematic and kinetic features of specific athletic movements, in order to find possible solutions for performance enhancement and, to a lesser extent, for injury prevention. Unfortunately, these surveys have often lacked either in providing a theoretical rationale or in presenting results that could be directly understood and practically used by trainers and athletes. The lack of methods and protocols allowing for in-field athlete evaluation as well as for accurate joint forces estimate represents, to date, the main reasons of this failure.

The main purpose of the present thesis is to provide a contribution towards the development of such methods, focusing, in particular, on sprint running evaluation. Two complementary approaches are adopted: a Low Resolution Approach, where the use of wearable inertial measurement units are exploited during different phases of sprint running, and a High Resolution Approach, where subject-specific, non-rigid constraints for knee joint kinematic modelling used in multi-body optimization techniques are defined.

LOW RESOLUTION APPROACH

The evaluation of athlete's performance is one of the main issues of coaching, as well as of sports biomechanical analysis. To this aim, in-field assessment of the athlete performance, without influencing or constraining athletes' activities, is now becoming mandatory.

Among the new wearable and lightweight technologies allowing for such assessment, inertial measurement units (IMUs) appear to be a good compromise between practicality and accuracy. These sensors combine three-axial accelerometers and gyroscopes, and, when a measure of a global reference frame is required, a magnetometer is also implemented. They allow data collection during unconstrained continuous movement over prolonged periods of time, potentially even during training and competition. Nevertheless, the extraction of movement-related information from the signal derived from IMUs can be strongly jeopardized by offset errors that rapidly accumulate over time (Woodman, 2007) and sensor wide oscillations caused by the inertia of soft tissues (de Leva & Cappozzo, 2006; Forner-Cordero et al., 2008).

The use of such sensors was explored in three studies aiming at estimating performance-correlated biomechanical variables during the different phases of sprint running (block-start, pick-up or acceleration, and maintenance phases). *Ad-hoc* methods aimed at reducing the above mentioned sources of error were defined. In particular, the trunk inclination and angular velocity, as well as the instantaneous horizontal velocity and displacement of the center of mass will be estimated during in-lab sprint running. The stride and stance durations will be assessed on-the-field during the maintenance phase.

Study one – Trunk inclination during the sprint start

The execution of the start is crucial in determining the performance during sprint running. Although trunk inclination is acknowledged to be a key element in moving from the crouch to the upright position, only few study focused on this parameter during the block start and the pick-up phases (Mero, Luhtanen, & Komi, 1983; Čoh, Jošt, Škof, Tomažin, & Dolenec, 1998; Slawinski, Bonnefoy, & Levêque, 2010).

The purpose of this study was to provide coaches with an instrument able to reliably estimate such parameter in-field. To this aim, the accuracy of an inertial measurement unit (IMU) in estimating its rotation about a local axis (referred to as “quality”) and the relationship between this rotation and trunk inclination in the progression plane (referred to as “consistency”) were assessed during block start and pick-up phases.

Five male sprinters performed four in-lab sprint starts. The block start phase and the first three steps of the pick-up phase were analysed. Data provided by an IMU (FreeSense, Sensorize Ltd, Italy) positioned on the trunk at L2 level were compared to reference stereophotogrammetric measurements. To reduce soft tissue oscillations, a memory foam material and an elastic belt were used. The trunk was modeled as a rigid segment joining C7 and the midpoint between the posterior superior iliac spines. The inclination of the unit (pitch angle: β) was estimated by combining the information provided by both the accelerometer, during the static phases of the movement, and the gyroscope, during the non-static phases. To improve the accuracy of such estimate, a Kalman algorithm (Kalman, 1960; Jurman, Jankovec, Kamnik, & Topic, 2007) was designed to automatically identify these static and non-static phases and to use a proper combination of the information provided by the two sensors. Root Mean Squared Errors, Pearson's correlation coefficient and Bland and Altman method (Nevill & Atkinson, 1997) were used to assess the quality and consistency of the estimates.

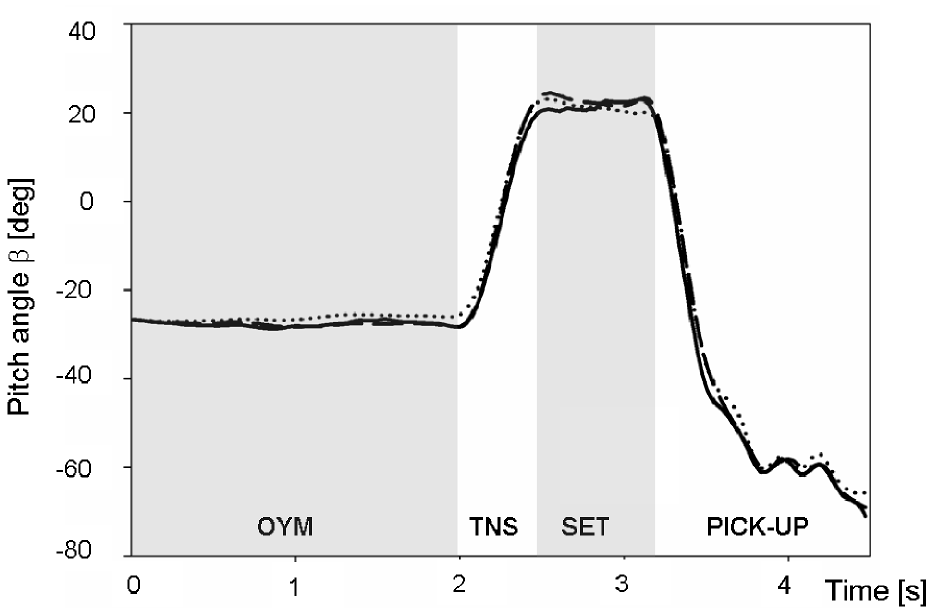


Figure 1: Typical pitch angles (β) for one trial as obtained from the IMU (solid line) and from the stereophotogrammetric system: IMU reference frame (dotted line) and trunk reference frame (dashed line). The different phases of the start are also indicated: OYM: "on your marks" position; TNS: transition phase; SET: "set" position, and PICK-UP: pick-up phase. The pitch angular displacement was considered to be zero when the unit was in a horizontal position; positive angles correspond to clockwise rotations.

The quality of the IMU estimates and their consistency with trunk inclination were high both in terms of curve similarity (correlation $r>0.99$) and bias (lower than 1 and 4 deg, respectively) (Fig. 1). The agreement between the unit and the trunk inclination, moreover, seems to support track and field coaches' approach in considering the trunk as a rigid segment. These results open a promising scenario for an accurate in-field use of IMUs for sprint start performance assessment.

Study two – Center of mass instantaneous velocity and displacement during the acceleration phase

When the focus is on sprint running performance determinants and limiting factors, biomechanical parameters as the stance duration, the step length and the center of mass (CoM) instantaneous velocity are crucial. Although these variables should be obtained by numerically integrating the acceleration signal provided, for instance, by inertial measurement units, in practice, offset errors that rapidly accumulate over time (Woodman, 2007) yield to unreliable velocity and, therefore, displacement.

The purpose of the study was to estimate the instantaneous horizontal velocity and displacement of a point approximating the CoM during sprint running by using a single inertial sensor. To this aim, a methodology for reducing the effects of the above mentioned errors was developed. Low frequency errors were compensated by reducing the numerical integration interval to the stance phase and by predicting the kinematics of the sensor during the flight phase. The initial conditions of the integration process were, then, cyclically determined.

Six sprinters performed three in-lab sprint runs, starting from a standing position. Due to limited laboratory volume only the first three steps were analysed. An IMU (MTx, Xsens, Netherlands) was positioned on the back trunk. Stereophotogrammetry and force platforms were used to validate final results. Reference and inertial sensors data were collected simultaneously at 100 samples per second. The instantaneous progression velocity and displacement were computed by numerical integration of the acceleration. The integration was limited to the stance phase only, to avoid the drift typical of the integration process. During the flight phase, the horizontal kinematics of the IMU was predicted using the ballistic law of motion; the velocity at the instant of take-off was the last value of the previously integrated acceleration. This procedure was reiterated for each step. The stance time (ST), CoM progression displacement (d) and the mean progression velocity (v) were estimated and compared with reference data. The method reliability was assessed by mean of multiple statistical tests (Multivariate ANOVA, Pearson's correlation coefficient, two-tailed paired t-test).

Results showed a high correlation (r>0.9) between IMU and reference estimates for each parameter (Fig. 2). No statistical differences were found between IMU and reference for v and ST.

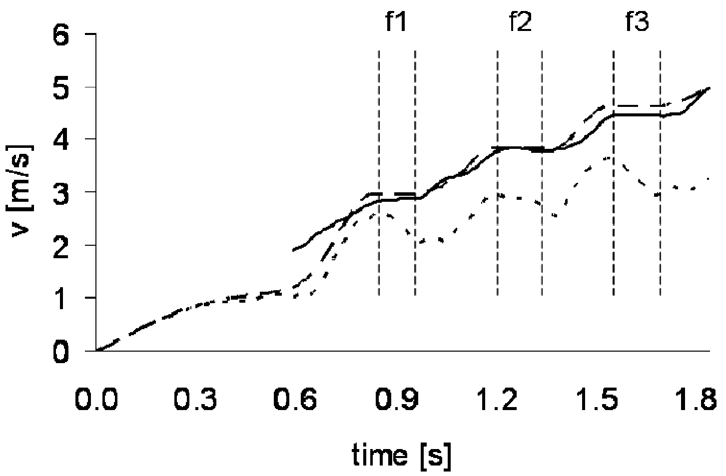


Figure 2: Instantaneous progression velocity as obtained by the reference measurements (solid line), computed by numerical integration of the acceleration for the whole duration of the task (dashed line) and with the algorithm proposed in this study (dashed line). Vertical dashed lines identify the flight phases (f1, f2 and f3).

The methodology proved to successfully compensate the numerical integration errors during in-lab non steady-state running. In-field validation is binding in order to provide track and field coaches with reliable and accurate information.

<u>Study three – Temporal parameters during the maintenance phase</u>

In the literature, walking and distance running temporal parameters have been generally determined by identifying mechanically-related features in the acceleration signal waveforms (Auvinet, Gloria, Renault, & Barrey, 2002; Kavanagh & Menz, 2008; Wixted, Billing, & James, 2010). Robustness and reliability of these temporal estimates, however, highly depends on the signal to noise ratio, especially unfavourable during sprint running analysis as the explosiveness of the task causes greater movements of the IMU relative to the underlying skeleton (Pain & Challis, 2006). For this reason, sprint running analysis is more challenging than walking or distance running.

The aim of the study was to identify recognizable and consistent features in the waveform of the signals supplied by a trunk mounted IMU, or thereof derived, for the estimation of stance duration during in-field sprint running.

Six amateur (group A) and five elite (group B) athletes performed three 60 m sprint runs on-the-track, starting from a standing position. Four steps at each athlete maximal speed were analysed. An IMU (FreeSense, Sensorize Ltd, Italy) containing a 3D accelerometer and gyroscope was positioned on the lower back trunk at L2 level with an *ad-hoc* elastic belt. To limit the unit oscillations relative to the underlying bone, a memory foam material was placed between the paravertebral muscles and the IMU (Fig. 3). Data provided by the IMU, acquired at 200 samples per second, were compared to reference forceplate and high-speed camera measurements. The magnitude of the acceleration (a) and angular velocity (ω) vectors as well as their 1st ($\dot{a}$ and $\dot{\omega}$) and 2nd ($\ddot{a}$ and $\ddot{\omega}$) wavelet-mediated derivatives were computed (Jianwen, Jing, & Jinhua, 2006). Features adequate for automatic detection of Foot-Strike (FS) and Foot-Off (FO) instants were identified and, thereafter, used to estimate the stance (d_{stance}) and stride (d_{stride}) durations. Repeated-measure ANOVA tests and Bland and Altman method (Nevill & Atkinson, 1997) were used to assess the accuracy of the estimates.

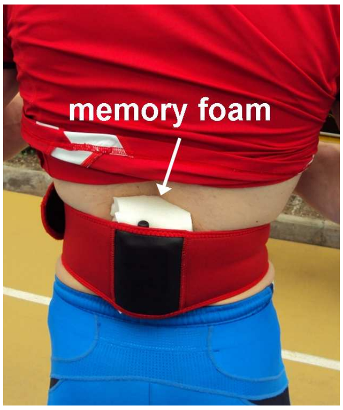


Figure 3: Belt and sensor unit location on the lower back trunk of an elite athlete of group B. Indication of the memory foam material location is also provided.

No repeatable and quantifiable features, adequate for automatic detection, were identified in either a or its derivatives. Conversely, the magnitude of the angular velocity signal was characterized by a consistent positive peak which occurred approximately at the end of each step cycle in both groups of athletes. This peak was clearly visible even by simple visual inspection of the signal and could be used to estimate d_{stride} (Fig. 4).

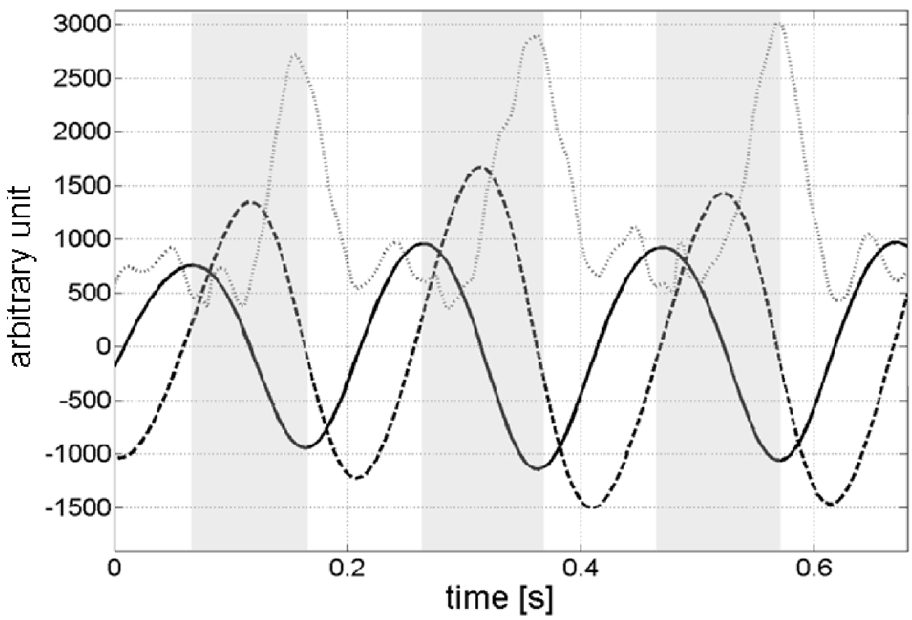


Figure 4: ω (grey dotted line), $\dot{\omega}$ (dashed line) and $\ddot{\omega}$ (solid line) with reference to a randomly chosen subject of amateur athletes (group A). Grey sections represent three consecutive stance phases.

The beginning and end of the stance were identified from positive and negative peaks on $\ddot{\omega}$ waveform (Fig. 4). These peaks were found to be consistently synchronized with FS and FO across steps, trials, subjects and groups. The mean of the absolute bias between the reference and the IMU estimates was found to be in the order of the temporal resolution of the IMU (0.005 s). It can be speculated that increasing that resolution may improve the final results. As track and field coaches' requirement is to obtain the stance time profile over time and during the whole race, future works will concern the validation of the method on different phases of the sprint run.

HIGH RESOLUTION APPROACH

A clear understanding of the definitive relationships between biomechanical measures and injury onset in sprint running would lead to better injury prevention strategies and would help track and field coaches to define effective training programs. In this respect, forces and force-related factors appear to be the prime agents that determine the likelihood and severity of injury. Epidemiological studies of sprint running injuries, in particular, found the knee to be the most frequent site of injuries (Brunet, Cook, Brinker, & Dickinson, 1990).

The estimate of knee joint forces during running would be, therefore, of great help for athletes, coaches and physicians.

The accuracy of inverse dynamics methods, often used in sports biomechanics to estimate internal and external forces, is affected by several sources of error. Among them, the motion of soft tissues relative to the underlying bones (soft tissue artefact – STA) is considered as the most important, particularly in sports applications. Different techniques have been proposed to compensate for STA. The multi-body optimization (MBO) method, in particular, aims at optimally estimating the location in space of a chain of bones interconnected in joints. Each joint is modelled by embedding specified constraints. To date, MBO has been performed using spherical, revolute or universal joint constraints (Lu & O'Connor, 1999; Andersen, Damsgaard, & Rasmussen, 2009), as well as using a parallel mechanisms (Duprey, Cheze, & Dumas, 2010). In light of recent results reported in the literature (Andersen, Damsgaard, & Rasmussen, 2009; Duprey, Cheze, & Dumas, 2010), the choice of joint constraints appears to be crucial.

Study four - Tibio-femoral joint constraints for multi-body optimization

To further improve the quality of knee joint models used in the MBO approach, the definition of non-rigid constraints which take into account the anatomy of the subject appear to be ideal. The aim of the study was to provide plausible, subject-specific values for the distances between the origin and insertion landmarks of the main knee ligaments (referred to as "ligament lengths"), during loaded continuous knee flexion-extension.

Two orthogonal digital radiographs of six knee specimens (femur, tibia, patella and fibula) were acquired using a low dosage X-ray system (EOS®, EOS-imaging, France). The 3D geometry of each specimen was then obtained by means of a reconstruction algorithm (Chaibi et al., 2011). The areas of origin and insertion of the anterior and posterior cruciate, lateral collateral, and deep and superficial bundles of the medial collateral ligaments (ACL, PCL, LCL, MCLdeep, MCLsup) were identified on femur and tibia templates using the mouse pointer by three expert orthopaedic surgeons (virtual palpation). Attachments sites were estimated for the six reconstructed knees by matching the bone templates to the low dosage stereoradiography images. Movement data of the specimens were obtained by means of a stereophotogrammetric system (Polaris, Nothern Digital Inc., Canada), using pins carrying a cluster of markers and inserted into the femur and the tibia. Data were, therefore, free from skin movement artefacts. For each knee, the centroids of the attachment areas of each ligament were determined and the Euclidean distance between the origin and insertion centroids computed (d_c). The impact of the inaccuracies associated with the virtual palpation was assessed performing a Monte Carlo simulation. The Euclidean distance between each possible couple of the points thus generated (100*100 couples) was also computed during knee motion (d_{MC}) (Fig. 5). Ligament length variations (Δd_{MC}) were then calculated relative to the distances at knee full extension and expressed as percentage of the latter value for each sampled knee flexion angle.

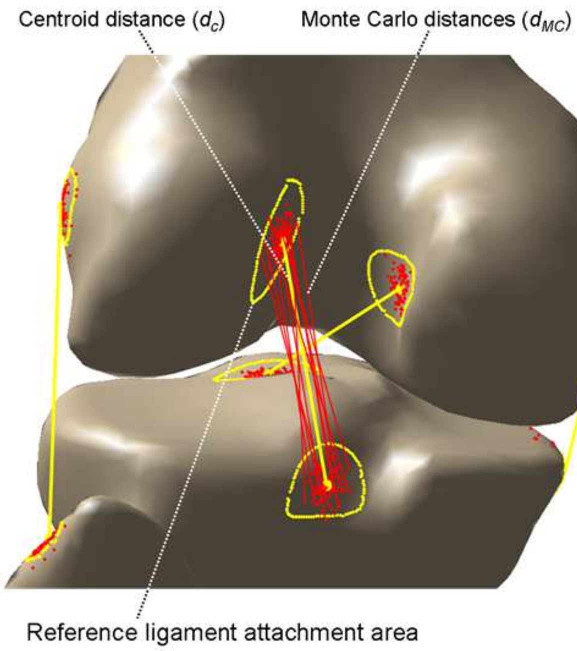
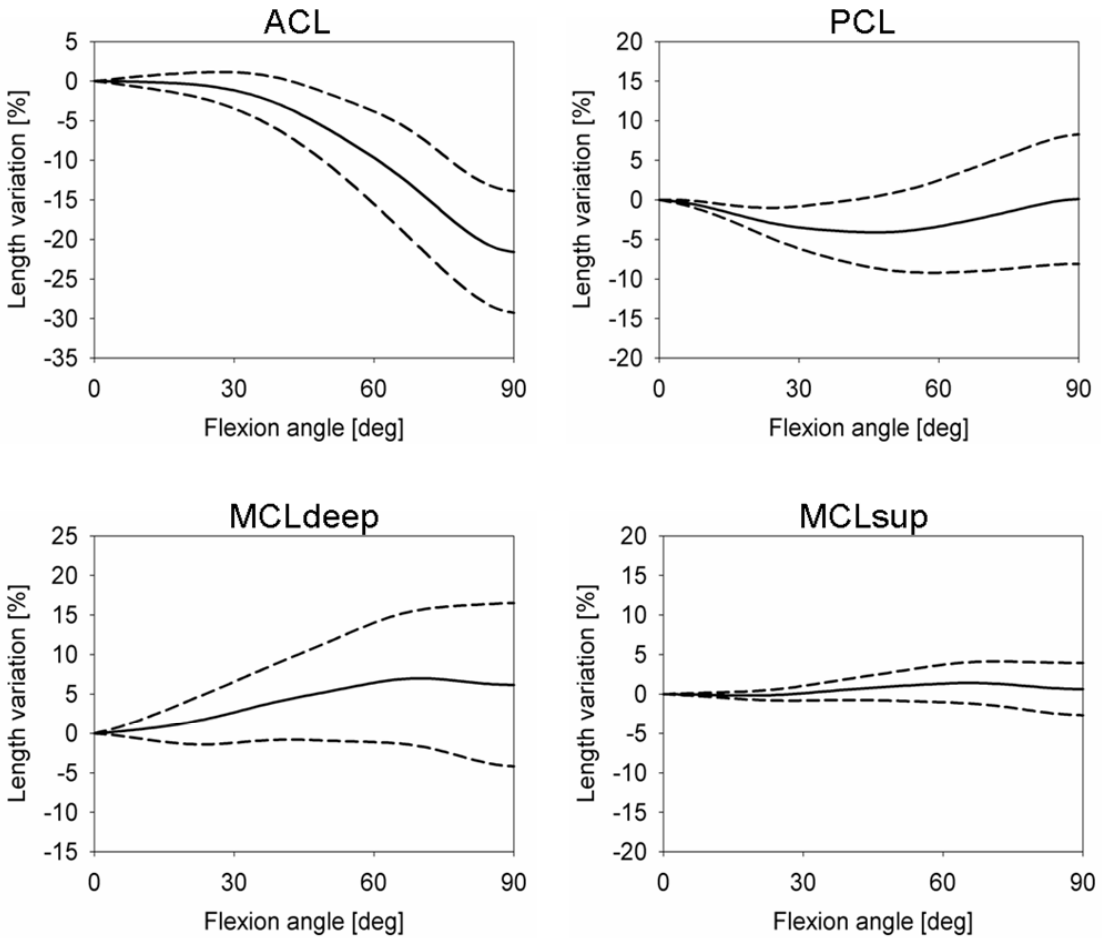


Figure 5: 3D digital model of one knee specimen: the ligament attachment areas and the tibio-femoral distances between the centroids (d_c) (yellow lines) as well as between selected Monte Carlo pairs (d_{MC}) (red lines) are depicted.

The mean and standard deviation (SD) curves of Δd_{MC} are shown in Fig. 6 for each ligament. To facilitate embedding this information in the knee kinematic model to be used in the MBO process, the mean of the Δd_{MC} curves vs flexion angle thus obtained, plus and minus one standard deviation, were fitted with a polynomial regression function of the fifth order. The ACL and LCL lengths were found to decrease, and the MCLdeep length to increase significantly during flexion, while PCL and MCLsup length trend of variation was concealed by the experimental indeterminacy. A mathematical model was provided that allowed the estimate of subject-specific ligament lengths as a function of knee flexion and that can be integrated in a multi-body optimization procedure. The efficacy of this model, as opposed to those already implemented, must be evaluated in terms of consequences on the estimate of joint kinetics, particularly when the inertial effects of soft tissue masses are involved.



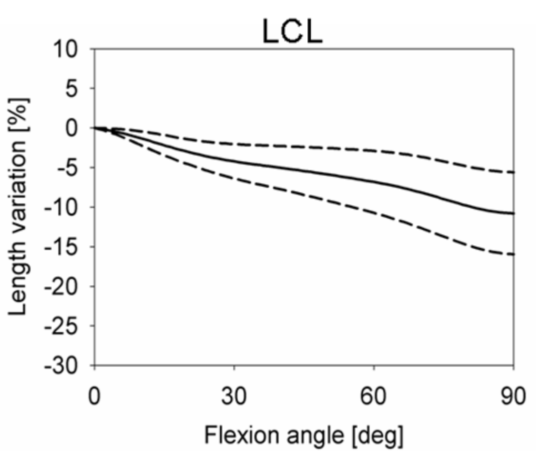


Figure 6: Distance variation patterns (mean ± 1SD) vs the knee flexion angle as obtained through the Monte Carlo simulation for each ligament, Δd_{MC}. Each variation is expressed as a percentage of the distance at knee maximal extension.

CONCLUSIONS

In the framework of sports biomechanics analysis, the results and the considerations carried out in the present thesis aim at providing a contribution towards the development of methods for in-field athlete evaluation as well as for accurate joint forces estimate.

Results about the Low Resolution Approach indicate that, due to their portability and inexpensiveness, inertial measurement units are a valid alternative to traditional laboratory-based instrumentations for in-field performance evaluation of sprint running. Using acceleration and angular velocity data, the following quantities were estimated: trunk inclination and angular velocity, instantaneous horizontal velocity and displacement of a point approximating the CoM, and stride and stance durations. In order to limit the motion of the soft tissue masses relative to the underlying bones, careful attention has to be paid to the location and method of fixation of the sensor. The use of memory foam materials and elastic belts appears to be effective. To limit the errors yielded by the unstable bias of the signal, the integration interval should be reduced, and boundary conditions used to cyclically correct the drift errors explored.

Results about the High Resolution Approach indicate that, in a kinematic model of the knee based on joint constraints, the length of the ACL, LCL and MCLdeep should be considered as variable during knee flexion. The length of the PCL and of the MCLsup was found to be highly dependent from the selected attachment sites. These ligaments could be, therefore, considered isometric during knee flexion. These results represent a first contribution to the definition of methods aiming at improving the accuracy of inverse dynamic estimates. On the other hand, as the MBO approach aims at providing an optimal estimate of the 3D position of a bone-chain, the reconstruction of the soft tissue movements may be attempted. A biodynamic model of the human body based on the reconstructed movement could then be developed. The combination of subject-specific constraint-based joint models with such biodynamic model appears to justify the investment of resources aimed at improving the MBO approach.

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CHAPTER 1

THEORETICAL BACKGROUND:

SPRINT RUNNING BIOMECHANICS

ABSTRACT

The continued increase in running popularity has prompted a comparable explosion of research in the last decades. This has been further potentiated by recent technical and technological advancements. The current state of knowledge about the major findings in sprint running biomechanics is presented and a brief overview of the current technologies used in the assessment of running is provided.

Many variables have been studied pertaining to the different phases of sprint running. Significant factors include: technique, electromyographic activity, kinematics and kinetics. Sprint technique has been analysed during the block start, acceleration and maintenance phases. The EMG activity pattern of the main muscles is described in the literature, but there is a need of further investigation, particularly for highly skilled sprinters. The reaction time of good athletes is short, but it does not correlate with performance levels. The force-power production and the force impulse during the block start phase are key factors in order to generate high velocity. Nevertheless, they proved to correlate with the incidence of knee-related injuries. During acceleration and maintenance phases, the reduction of the horizontal braking forces and the maximisation of the propulsive forces are crucial in order not to decrease velocity. Leg and vertical stiffness are sensitive parameters for the optimization of performance and, at the same time, for the reduction of injury risk. Several external factors, as footwear, ground reaction surface and air resistance, may influence the athlete's technique and performance. Efficient sprint running requires an optimal combination between the examined biomechanical variables and such factors.

Interestingly, while a large number of studies focused the determinants of the performance, there is a general paucity of scientific works showing definitive relationships between either anatomical factors and injury, or biomechanical measures and injury during sprint running.

As concerns technologies and methods for sprint running analysis, although traditional measurement devices such as motion capture systems, force plates, and electromyography are considered as the most accurate methods, they suffer from

limitations, such as expense and lack of portability. Recent technological advances have made available more viable options such as accelerometers, electrogoniometers, gyroscopes, and in-shoe pressure sensors. Combined with wireless technology and/or data loggers, they appear to be an affordable, lightweight alternative to running analysis, allowing data collection over prolonged periods of time in almost any environment.

KEYWORDS: Sprint running; Biomechanical variables; Methods; Technology; State of the art.

1.1 INTRODUCTION

It is mandatory to start an issue about running biomechanics in the following classical way: human running is characterised by a phase of the locomotor activity during which the body is not in contact with the ground. This means that the demarcation between walking and running occurs when periods of double support during the foot-ground contact phase (or stance phase) of the gait cycle (both feet are simultaneously in contact with the ground) give way to two periods of double float at the beginning and the end of the swing phase of gait (neither foot is touching the ground) (Fig. 1). This is, in broad terms, the definition provided by E. J. Marey following his experimental acquisitions with his "chassure dynamographique".

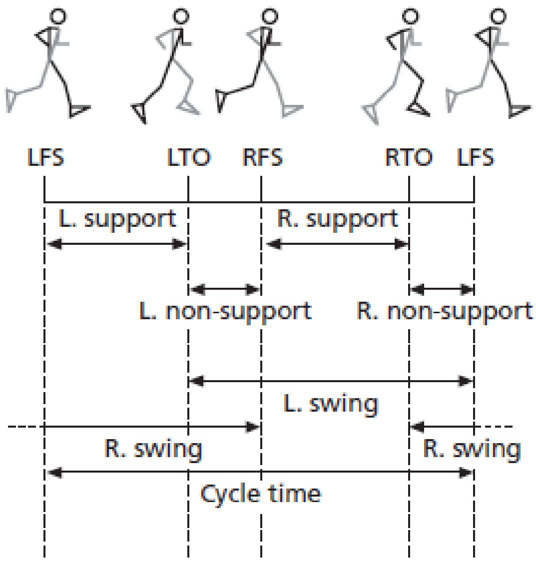


Figure 1 (1.1): Left and right foot temporal parameters (foot-strike: LFS, RFS, and toe-off: LFO, RFO) during running (from Zernicke & Whiting, 2000).

It is evident that running, referred to as a specific motor paradigm, is legitimated by the fact that it allows for higher progression speed with respect to walking and race-walking. Generally, as speed further increases, initial contact changes from being on the hindfoot to the forefoot. This typically marks the distinction between running and sprinting. The higher the progression speed, the shorter the duration of the stance phase and the longer the flight phase time. Novacheck et al. (1998) reported that, during running, stance and flight phase durations are respectively about 40% and 60% of the step cycle, while in elite sprinters, duration can reach 20% and 80% respectively. For distance running the body is moved at a controlled rate in relation to the energy demand of the race. For sprinting, on the other hand, the body and its segments are moved as rapidly as possible throughout the entire race. As an example of the different movement strategies adopted during running and sprinting, ankle, knee and hip joint kinematics at different progression speed are reported in Fig. 2.

The difference between running and sprinting is in the goal to be achieved (Novacheck, 1998). Running is performed over longer distances, for endurance, and primarily with aerobic metabolism. Jogging, road racing, and marathons are examples. Sprinting activities are done over shorter distances and at faster speeds, with the goal of covering a relatively short distance in the shortest period of time possible without the need to preserve aerobic metabolism (Novacheck, 1998).

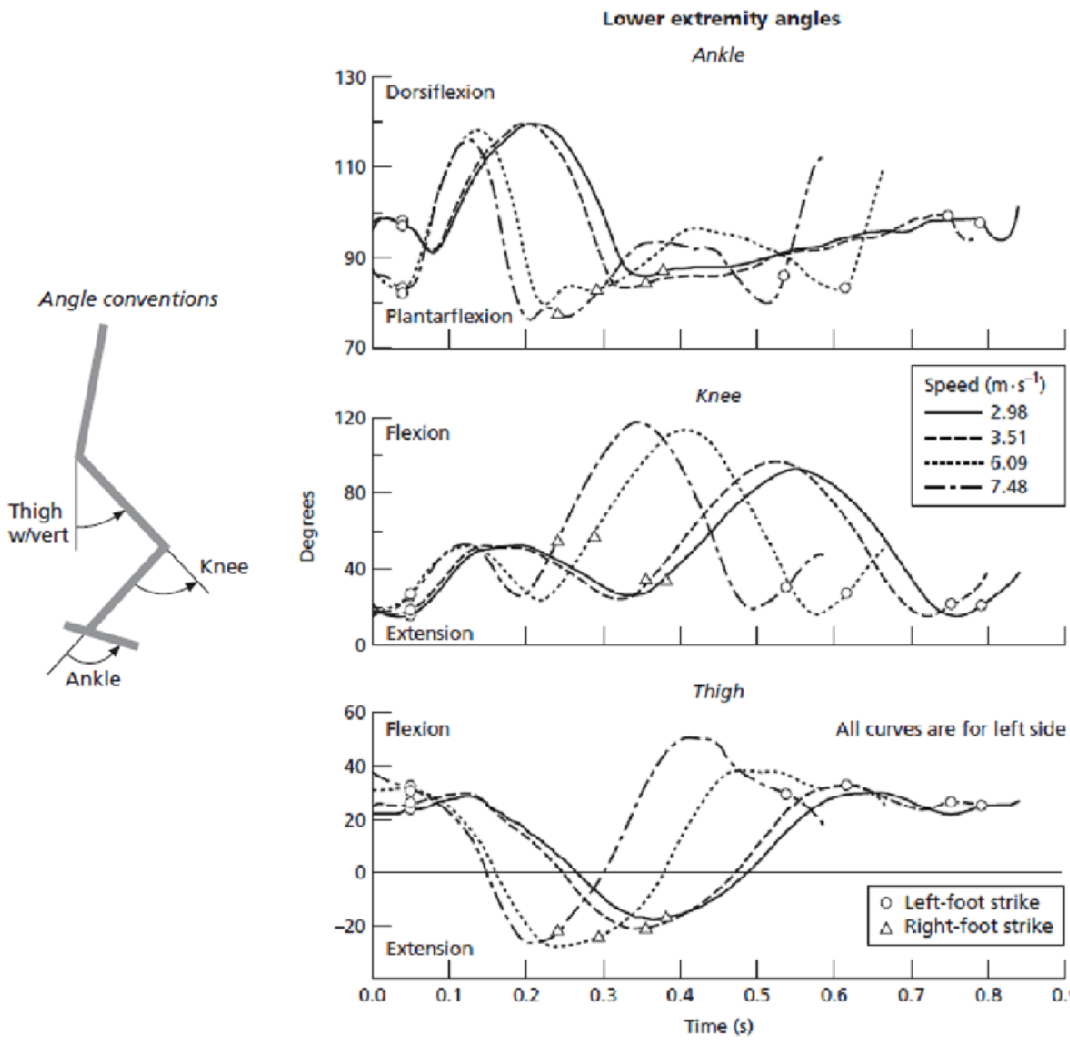


Figure 2 (1.1): Ankle, knee and hip motion time-curve throughout a running cycle at four different running speeds: an example runner. Angle conventions are also indicated (adapted from Zernicke & Whiting, 2000).

Rapid movements of the body from one place to another are required in many sports activities and, particularly, in track and field sprinting events, where athletes' objective is simply to cover a given distance (either on the flat or over obstacles) in the least possible time (Hay, 1993). Among running track sprinting events, the most studied and considered is the 100 m race. It is the shortest outdoor sprint race distance in athletics and it is often used as a paradigm to describe and analyse the different phases characterising sprint running and the relevant biomechanical factors influencing the athlete performance.

1.2 SPRINT RUNNING BIOMECHANICS: PERFORMANCE AND INJURY-RELATED VARIABLES

Early studies on the velocity-time curve in sprint running were concluded by Hill (1927). Since then, there has been a great deal of research (Volkov & Lapin, 1979; Tellez & Doolittle, 1984), including the mathematical representation of such curve (Henry, 1954; Morton, 1985). A typical speed-time curve, as measured and modelled by Morin et al. (2006), is reported in Fig. 3.

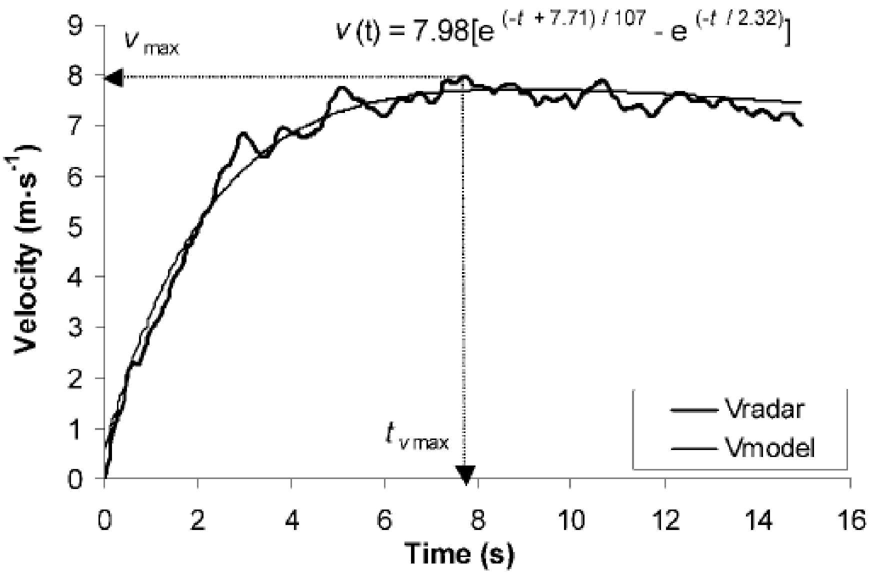


Figure 3 (1.2): Typical speed-time curve measured by the radar (bold line) and modelled by the biexponential equation (gray line). In this example, $v$max =7.98m·s−1, $t$vmax = 7.71 s, and the time constants for acceleration and deceleration are respectively τ1 = 2.32 s and τ2 =107s (from Morin, Jeannin, Chevallier, & Belli, 2006).

Although no general consensus has still been reached by the scientific community, as well as by track and field coaches, about the number and type of phases in which a 100 m sprint race should be divided (Jones, Bezodis, & Thompson, 2009), in the present review four different phases will be considered: the block-start, the acceleration or pick-up phase, the maintenance and the final deceleration. There are many factors that affect the duration of each of theses phases. Internal or personal factors, such as motivation, technique, fitness and fatigue, as well as external aspects, like strength and direction of wind, air temperature, and texture or hardness of the track surface (Nigg & Yeadon, 1987; Stafilidis & Arampatzis, 2007). Their time-duration is, therefore, highly variable even when considering multiple races of the same athlete and should not be used as a criterion in the identification of different phases. From Jones et al., (2009), the most used criterion appears to be the effectiveness of the coaching activity, in terms of giving to the athletes the necessary level of specificity in the instruction and feedback. In this respect, the start phase is defined as ranging from when the athlete obtained a “Set” position in the blocks to the point when the front foot broke contact with the block. The acceleration/pick-up phase is defined as being from when the athlete’s front foot left the block to the point when he/she attained an upright sprinting position. The maintenance phase is seen as the phase in which the athlete is able to maintain his/her velocity almost constant. Finally, the deceleration phase is defined as the remainder of the race; that is, from when the athlete’s velocity starts to decrease to when the finishing line is crossed (Mero, Komi, & Gregor, 1992).

In the following sections, the execution technique, the pattern of muscular activation (electromyography – EMG) as well as the main kinematic and kinetic variables for each of the four identified phases will be discussed.

1.2.1 Block start phase

The block start phase refers to the time when the sprinter is in contact with the starting blocks. Blocks have been regularly used in track competitions under the International Amateur Athletic Federation (IAAF) rules since 1948, the year of London Olympic Games.

The block start phase starts when the track judge gives the “On your marks” command and ends with the athlete block clearing. After the “On your marks” command, the judge gives the “Set” order and finally a gun is fired (or else there is a final “Go” command by the judge) (Fig. 4). When the athlete hears the initial command, "On your marks", he/she moves forward and adopts a position with the hands shoulder width apart and just behind the starting line. The feet are in contact with the starting blocks and the knee of the rear leg is in contact with the track. On hearing the command "Set" the athlete raises the knee of the rear leg off the ground and thereby elevates the hips and shifts the body centre of mass (CoM) up and out. Then on the command "Go" or when the gun is fired the athlete reacts by lifting the hands from the track, swinging the arms vigorously and driving with both legs off the blocks and into the first running strides (Fig. 4).

Figure 4 (1.2): The action sequence during the block start phase (adapted from Hay, 1993).

The purpose of the block start is to facilitate an efficient displacement of the athlete in the direction of the run. The main objectives of the athlete during this phase can be summarised as follows (Tellez & Doolittle, 1984):

- To establish a balanced position on the blocks.
- To obtain a body position with the CoM as high as it is practical and slightly forward of the base of support.
- To apply a force against the blocks whose line of action goes through the ankle, knee and hip joints, the centre of the trunk and of the head.
- To apply this force against the blocks and through the body at an angle of approximately 45°.
- To clear the blocks with the greatest possible velocity.

EMG activity

The first aspect to be considered when analysing the block start phase is the reaction time. It has been defined as the time that elapses between the sound of the starter's gun and the moment the athlete is able to exert a certain pressure against the starting blocks. Reaction time measurement currently includes the time that it takes for the sound of the gun to reach the athlete, the time it takes for the athlete to react to the gunshot and the mechanical delay of measurements inherent in the starting blocks.

An attempt has been made to separate premotor time and motor time components in the block start phase (Mero & Komi, 1990). The former is defined as the time from the gun signal until the onset of EMG activity in skeletal muscle. Motor time is the delay between the onset of electrical activity and force production by the muscles. EMG results (Mero & Komi, 1990) showed that total reaction time can be effectively divided into premotor and motor time. However, electrical activity in some muscles started to increase after total reaction time as a result of the multi-joint nature of the sprint start movement. It is clear that, after the gun signal, leg extensor muscles must contribute maximally to the production of force and ultimately to the running velocity. The faster the electrical activity begins in every muscle, the faster the athlete can be in maximising the neuromuscular performance. For improving the start action, it is desirable that all extensor muscles are activated before any force can be detected against the blocks.

Mero and Komi (1990) used a force threshold of 10% from the maximal horizontal force production as a measure of reaction time. Total reaction time was on average 120 ms, which was the minimal reaction time for a valid start in the Rome World Championships in 1987. In fact, no definitive study exists which could be used to establish a minimum reaction time to define a false start. For comparison of reaction times to be used, uniform conditions for measurement must be established.

The main conclusions regarding reaction time during the block start phase are (Moravec et al., 1988):

1. In identical events the average reaction times of women are longer than those of men;
2. Reaction times grow in proportion to the length of race distance;

3. Reaction time does not correlate with the performance levels, thus meaning that other parameters, as acceleration or maximum speed, may be more important than reaction time to final race performance.

General muscle activation patterns during the block start phase showed considerable individual variances (Mero & Komi, 1990) (Fig. 5). Results provided support for the concept that the gluteus maximus muscle is very active at the beginning of force production, while the gastroctemio muscle is the first muscle to become activated. The biceps femoris, as well, proved to be a very important muscle during the early stages of the sprint start. The duration of the force production by the front leg is nearly the same as that during the total block phase, because the front leg produces force from the beginning of the total force production to the end of the block phase (Baumann, 1976).

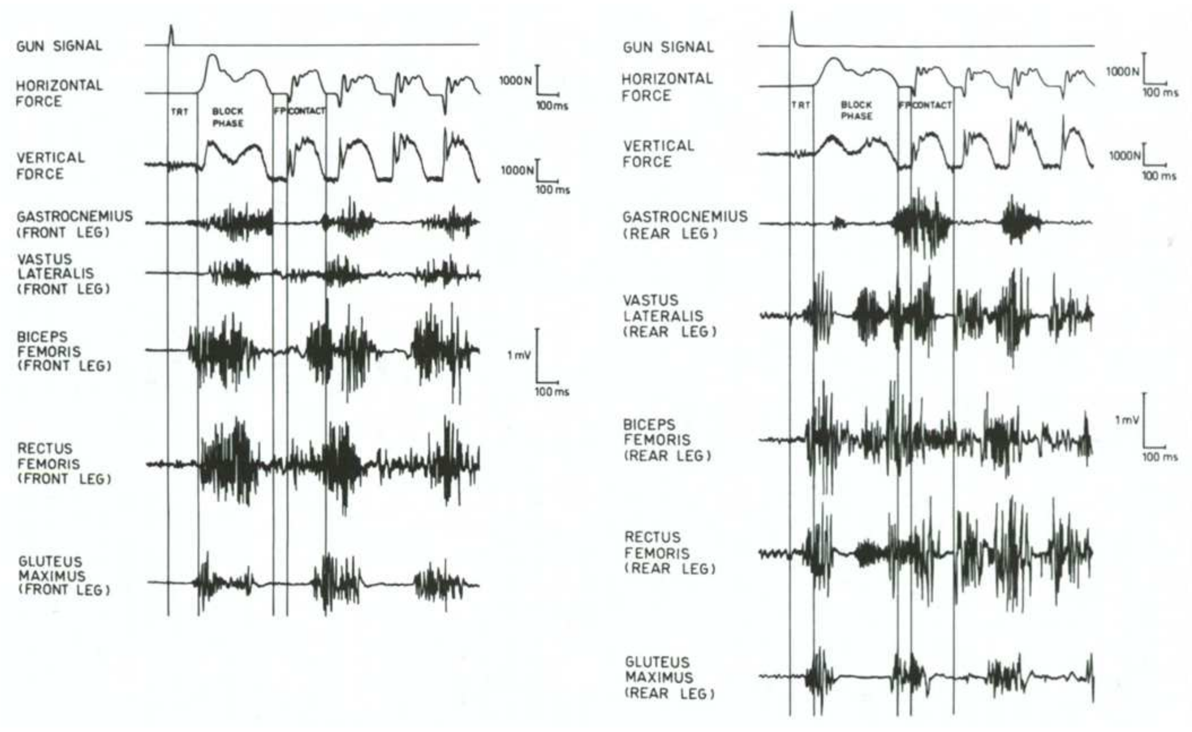


Figure 5 (1.2): Raw electromyographs of selected muscles in front and rear legs during maximal block start of one subject. TRT = total reaction time; FP = flight phase. The ground reaction force (horizontal and vertical components) is also displayed (from Mero & Komi, 1990).

Kinematics and kinetics

Many kinematic and kinetic variables have been studied pertaining the block start phase, over the past decades (Payne & Blader, 1971; Baumann, 1976;

Mero, Luhtanen, & Komi, 1983; Cappozzo, Gazzani, & Massacesi, 1989; Schot & Knutzen, 1992; Fortier, Basset, Mbourou, Faverial, & Teasdale, 2005; Čoh, et al., 2006; Slawinski, Bonnefoy, & Levêque, 2010; Slawinski et al., 2010). Different biomechanical variables were obtained and were shown to contribute to a fast start technique (Harland & Steele, 1997). In the following section, the main results about biomechanical variables are discussed.

- Inter-block distance: a medium block spacing (as opposed to the bunched or the elongated one), combined with the hips raised high in the set position, was theorised to enable sprinters to utilise more completely an extensor reflex of the muscle groups relevant to sprint starting (Schot & Knutzen, 1992). Furthermore, the medium starting position produced the fastest acceleration than enables both a powerful and quick recovery of the rear lower extremity (Čoh, et al., 2006).
- Block inclination: sprint start performance was shown to improve when decreasing block inclination. This improvement was attributed to an increased contribution of the medial gastrocnemius muscle during the eccentric and concentric phases of calf muscle contraction due to an earlier onset. This increased contribution appeared to be the result of progressive lengthening of the soleus and gastrocnemius muscles in the set position as the front block inclination decreased. Therefore, during the subsequent stretch-shorten cycle, force production was improved, more effectively providing and elastic contribution to the speed of muscle shortening (Mero, & Komi, 1990).
- Trunk and knee alignment during the “Set” position: an optimal “Set” position was shown to exist for highly skilled sprinters irrespective of variations in body structure. In particular, the stronger the sprinter, the more acute the joint angles can become. That is, stronger sprinters can use a greater range of joint extension to gain greater velocity when leaving the blocks (Mero, Luhtanen, & Komi, 1983; Slawinski, Bonnefoy, & Levêque, 2010). Knee angles during the “Set” position for elite and non-elite athletes are reported in Tab. 1.
- Hip joint alignment during the “Set” position: hip joint angles in the “Set” position have been found to significantly differ between good and average sprinters. In particular, skilled athletes reported lower joint angle values both for the front and the rear legs (41 and 80 deg respectively, against 52 and 89

deg for average athletes) (Mero, Luhtanen, & Komi, 1983; Slawinski, Bonnefoy, & Levêque, 2010). These findings suggested that more skilled sprinters placed their hip extensors muscles on a greater stretch than their less skilled counterparts.

- CoM position during the "Set" position: positioning the CoM as close as possible to the start line in the antero-posterior direction was suggested to be important in creating a good start, as it contributes to reach a position of maximum instability and it moves the athlete prospectively closer to the finish line, thus reducing the distance the sprinter must accomplish (Slawinski, Bonnefoy, & Levêque, 2010). However, too pronounced forward trunk lean in the "Set" position has to be avoided, as it excessively loads the hands (Tellez & Doolittle, 1984).

Set position	Elite sprinters (± *SD*)	Well-trained sprinters (± *SD*)
XCM (cm)	−22.9 ± 1.5	−27.81 ± 2.8†
YCM (cm)	65.7 ± 3.8	62.6 ± 3.9
Xshoulder (cm)	10.7 ± 2.7	4.0 ± 5.5†
Front knee angle (°)	110.7 ± 9.3	106.1 ± 13.7
Rear knee angle (°)	135.5 ± 11.4	117.3 ± 10.1†

Table 1 (1.2): Front and rear knee angles for elite and well-trained sprinters during the "Set" position. The antero-posterior (XCM) and vertical (VCM) components of the CoM position with respect to the start line, together with the antero-posterior position of the shoulders (Xshoulder: midpoint of the line joining the right and the left acromions) are also provided (from Slawinski, Bonnefoy, & Levêque, 2010).

- Block time, velocity and acceleration: block time is defined as the time from the beginning of force production, with either foot, to the point where no further force production occurs against the blocks (i.e. block clearing) (Mero, Luhtanen, & Komi, 1983). From published data (Harland & Steele, 1997), it appears that more skilled sprinters exhibited shorter block times compared with their less skilled counterparts. As concerns resultant velocity and acceleration of the sprinter at loss of foot contact with the block (usually referred to as block or leaving velocity and acceleration), it is well documented how the ability of an athlete to leave the blocks at a high velocity generally increases with an increase in his/her force application on the blocks (Baumann, 1976; Cousins & Dyson, 2004; Mero & Komi, 1990; Slawinski, Bonnefoy, & Levêque, 2010; Čoh, et al., 2006).

- Peak-force production: faster sprinters have been characterised as being capable of more adequate propulsion on the rear block during their starts compared with slower sprinters (Slawinski, Bonnefoy, & Levêque, 2010). Moreover, skilled sprinters generally apply lower peak force on the rear block compared to the front block, with the rear block forces being exerted more rapidly. Fig. 6 reported the force time-curves measured on the rear and front blocks during the sprint start of a middle-level athlete (Payne & Blader, 1971).

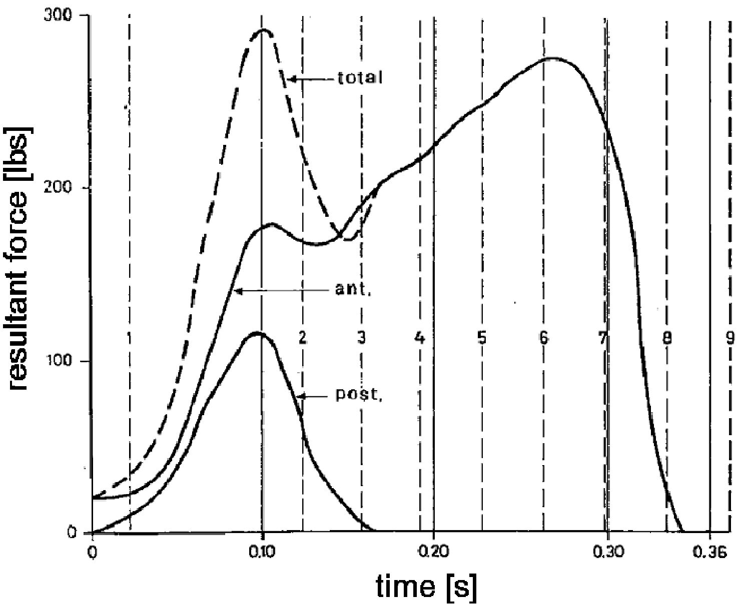


Figure 6 (1.2): Resultant forces measured on the front (ant.) and on the rear (post.) starting blocks, as well as measured on both blocks (total) (adapted from Payne & Blader, 1971).

- Direction of force application: it has been claimed that a good start is characterised by the exertion of high forces in the horizontal direction (Baumann, 1976). The angle between the horizontal and the line joining the CoM to the front toe at the loss of front contact has been reported to range from 32 to 42 deg for skilled sprinters (Mero, Luhtanen, & Komi, 1983). Angles of force application relative to the horizontal have been reported to vary form 43 deg at loss of block contact to 50 deg at toe-off of the first step (Mero, Luhtanen, & Komi, 1983).

- <u>Force impulse</u>: impulse incorporates both block force and block time and it is representative of the average amount of force serving to propel the sprinter and the time over which this force acts. For skilled sprinters horizontal and vertical impulses have been reported to range from 233 to 234 Ns and 172 to 231 Ns, respectively (Mero, Luhtanen, & Komi, 1983; Slawinski, Bonnefoy, & Levêque, 2010). Baumann et al. (1976) also reported that faster sprinters were able to exert a greater impulse in the horizontal direction (263 Ns) than less skilled sprinters (214 Ns). As block time was not significantly different between these groups, the greater impulse exhibited by the elite sprinters was created by a greater average force production. Similar results have been reported recently by Slawinski et al. (2010).

1.2.2 Acceleration or pick-up phase

After the block clearing, the runner accelerates by increasing stride length and stride rate. The pick-up phase ranges from the block clearing to the instant of time in which the athlete attains an upright sprinting position. Its length is about 30 to 50 m in top sprinters during a 100 m race (Volkov & Lapin, 1979; Moravec et al., 1988). Two key aspects have been identified in the acceleration phase: arm action and leg extension (Jones, et al., 2009). As concerns the arm action, Thomson et al. (2009) identified how previous research (Hinrichs, Cavenagh, & Williams, 1987; Mann, Kotmel, & Herman, 2008) documented the arms' balancing function in relation to the motion of the legs while sprinting. Nevertheless, up to now, no general consensus has been displayed about the amount of elbow and shoulder flexion and extension and no work seems to have specifically analysed the action of the arms within the pick-up phase. Leg extension refers to the hip and knee joints being fully extended prior to the athlete taking-off from each step in order to maximise the force exerted onto the running track. With this respect, it is necessary for both legs to have the same behaviour, symmetrical but alternate (Collier, 2002; Hunter, et al., 2004b). Two more aspects have been reported to be important during the acceleration phase: first contact times: Coh and Tomazin (2006) noted that contact phases become shorter and flight phases longer as the athlete progresses from the starting blocks; second the athlete's posture: due to the development of running velocity and the subsequent dynamic changes in running technique, the athletes' ability to maintain their dynamic posture as opposed to the static posture of the sprint start appears to be crucial (Jones, et al., 2009).

EMG activity

Integrated EMG activity during acceleration has been reported by Mero and Peltola (1989). In that study two male sprinters ran a 100 m simulated race and, in the acceleration phase, there was a 4.8% higher maximal integrated EMG activity during contact than in the maximum constant speed phase. This may imply that neural activation of sprinters achieves its maximum in the acceleration phase.

Kinematics and kinetics

The main kinematic and kinetic parameters which have been investigated during the pick-up phases are hereafter discussed.

- Stance and flight phase duration: mean stance times during the acceleration phase for elite male sprinters have been shown to range from 0.160 s to 0.194 s for the first ground contact out of the blocks (Atwater, 1982; Mero & Komi, 1990; Čoh, et al., 2006), and from 0.150 to 0.181 s for the second ground contact (Atwater, 1982; Čoh, et al., 2006). Flight phases are characterised by lower durations: from 0.06 to 0.07 s for the first flight and from 0.044 to 0.090 s for the second flight phase (Atwater, 1982; Čoh, et al., 2006). Conversely, after the first steps, stance times tend to decrease while accelerating, and flight times to increase (Zernicke & Whiting, 2000).
- Step length: this is defined as the distance between the first contact point of one foot to the first contact point of the controlateral foot (differently to "stride length", which is the distance between the first contact point of one foot to the first contact point of the same foot, like for example: right-left-right foot contacts). Increasing the length of the first step out of the blocks has been advocated as part of an optimal start (Čoh, et al., 2006). Nevertheless, anterior braking forces associated with the CoM being ahead of the first ground contact point were not significantly higher that that created with shorter steps (Schot & Knutzen, 1992). Moreover, an overly long first step may retard progress of the

sprinter. It has also been shown how the degree of forward lean in the “Set” position had a varying effect on first step length (Schot & Knutzen, 1992). Tab. 2 reports the step length, as well as the stance and flight times for the first ten steps of an elite sprinter during the acceleration phase (Čoh, et al., 2006).

Variable	Unit	1	2	3	4	5	AS SD
20M SPRINT	s	**3.08**	**2.98**	**3.07**	**3.03**	**3.19**	**3.07 ± 0.08**
Step number	n	12	12	12	12	12	12.00 ± 0.00
Step frequency	Hz	4.5	4.4	4.6	4.6	4.6	4.54 ± 0.09
Step length	cm	165	166	162	163	163	163.80 ± 1.64
Ground contact time	ms	125	126	126	126	129	126.40 ± 1.52
Flight time	ms	96	100	93	95	87	94.20 ± 4.76
Activity index – contact/flight		1.30	1.26	1.35	1.32	1.48	1.34 ± 0.11
STEP ONE							
Length	cm	103	103	103	103	106	103.60 ± 1.34
Ground contact time	ms	172	178	184	167	185	177.20 ± 7.73
Flight time	ms	62	37	56	55	43	50.60 ±10.26
STEP TWO							
Length	cm	99	105	108	102	105	103.80 ± 3.42
Ground contact time	ms	142	179	154	154	166	159.00 ± 9,04
Flight time	ms	86	80	80	92	74	82.40 ± 6.84
STEP THREE							
Length	cm	133	136	130	130	133	132.40 ± 2.51
Ground contact time	ms	141	129	135	129	148	136.40 ± 8.17
Flight time	ms	80	92	86	80	73	82.20 ± 7.16
STEP FOUR							
Step length	cm	136	140	143	136	133	137.60 ± 3.91
Ground contact time	ms	130	130	130	136	130	131.20± 2.68
Flight time	ms	110	92	104	92	98	99.20 ± 7.82
STEP FIVE							
Step length	cm	158	155	158	158	158	157.40 ± 1.34
Ground contact time	ms	111	129	123	123	117	120.60 ± 6.84
Flight time	ms	86	86	93	87	92	88.80 ± 3.42
STEP SIX							
Step length	cm	155	164	164	161	158	160.40 ±3.94
Ground contact time	ms	117	130	129	123	117	123.20 ± 6.26
Flight time	ms	99	98	92	98	105	98.40 ± 4.62
STEP SEVEN							
Step length	cm	171	177	180	174	177	175.80 ± 3.42
Ground contact time	ms	129	117	117	123	117	120.60 ± 5.37
Flight time	ms	86	111	111	93	105	101.20 ± 11.23
STEP EIGHT							
Step length	cm	177	192	186	183	183	184.20 ± 5.45
Ground contact time	ms	117	111	105	117	110	112.00 ± 5.10
Flight time	ms	111	117	117	104	111	112.00 ± 0.09
STEP NINE							
Step length	cm	186	189	192	189	189	189.00 ± 2.12
Ground contact time	ms	99	98	104	111	105	103.40 ± 5.22
Flight time	ms	92	111	111	105	105	104.80 ± 7.76
STEP TEN							
Step length	cm	186	196	199	196	196	194.60 ± 4.98
Ground contact time	ms	117	105	111	110	110	110.60 ± 4.28
Flight time	ms	104	123	123	111	117	115.60 ± 8.17

Table 2 (1.2): Step length and ground contact and flight times of a professional sprinter during the first ten steps of the acceleration phase (from Čoh, et al., 2006).

- Center of gravity (CoG) position: once the sprinter has started to leave the blocks, his/her task is to prepare for the subsequent ground contacts that he/she will make to assume maximal sprint velocity. If the horizontal position of the first foot to contact the ground after block clearing is posterior to the CoG along the antero-posterior direction, the sprinter is immediately able to maximize horizontal force production. Coh and Tomazin (2006) identified the position of the foot contact as being crucial to the successful execution of the pick-up phase, specifically minimizing braking forces during the first step. Indeed, the position of the CoG with respect to the first contact point on the ground changes during the first few strides. At the beginning of the first two stance phases, it is ahead of the foot-ground contact point. By the beginning of the third stance phase, the CoG is already behind the contact point (Mero, Luhtanen, & Komi, 1983; Slawinski, Bonnefoy, & Levêque, 2010).
- CoM vertical displacement: during the initial ground contact phases following block clearance, the CoM falls vertically. This vertical displacement reduces the step rate because of increasing ground contact time and, in turn, reduces the running velocity (Mero, Luhtanen, & Komi, 1983). Elite sprinters have been found to exhibit a reduced CoM negative vertical displacement (0.017 ± 0.016 m) during the eccentric phase of the first stance phase compared with slower sprinters (0.027 ± 0.014 m).
- CoM horizontal velocity: Fig. 7 reports the time-curve of the CoM horizontal velocity of elite and well-trained sprinters during the first two steps of the acceleration phase (Slawinski, Bonnefoy, & Levêque, 2010). Similarly, Mero et al. (1983) reported a mean horizontal velocity of the CoM of 5.7 $m \cdot s^{-1}$ at toe-off of the second post-block step for skilled male sprinters.
- Trunk alignment: it has been reported to be approximately 45 deg relative to the horizontal at loss of contact with the front block (Atwater, 1982; Van Coppenolle, et al., 1990; Slawinski, Bonnefoy, & Levêque, 2010). As concerns the trunk orientation during the pick-up phase, however, no general consensus on the best technique has been reached yet and no work seems to have specifically analysed such variables within this phase.

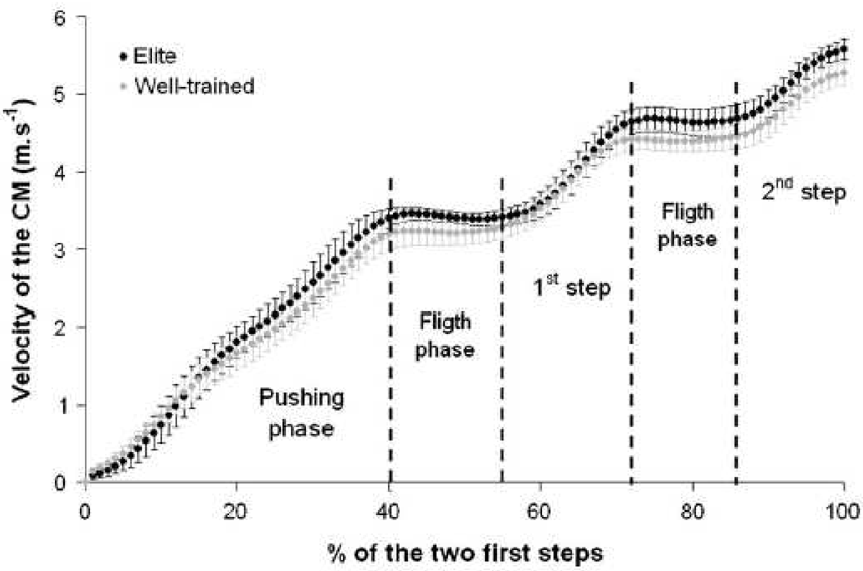


Figure 7 (1.2): Evolution of the CoM horizontal velocity of elite and well-trained athletes during the pushing phase and the first two steps of the acceleration phase (from Slawinski, Bonnefoy, & Levêque, 2010).

- Force production: despite the forward position of the CoG with respect to the first ground-contact point, a negative horizontal force is observed during the first step, probably caused by the leg moving forwards (Schot & Knutzen, 1992). This suggests that in sprint running, all stance phases are characterised by braking and propulsive components of the ground reaction force (GRF) (Hunter, et al., 2005), although the ratios are different according to the race phase. In particular, average horizontal forces in the first portion of the race are considerably larger (526 N) with respect to their braking counterparts (153 N) (Mero, Komi, & Gregor, 1992). Interestingly, Mero et al. (1992) reported that the horizontal propulsive force exerted during the first step after block clearing was 46% greater than the same force generated during contact at maximum velocity. This result highlights the need for the athlete to use a high level of concentric strength during the acceleration phase.

1.2.3 Maintenance phase

Professional track and field coaches describe a good running technique as the sum of four crucial aspects. First, the ability of the athlete in maintaining a relatively high CoM, with a slight anterior tilt of the pelvis during the maintenance phase of the race. Second, the needs of performing a wide hip flexion during each flight phase, trying to reach what coaches call “the high hip position” (Collier, 2002). Third, the importance of arm action, with the athlete swinging the arms exclusively in the progression plane, not across the body and with the elbow angle maintained close to 90 deg of flexion. The movements of both arms should be the same, although opposite in direction and they should be corresponding and complementary (Čoh, et al., 2006; Jones, et al., 2009). During the maintenance phase, the arms should work as ‘balancing factor’, by providing lift and promoting a more constant horizontal velocity for the runner (Hinrichs, Cavenagh, & Williams, 1987; Jones, et al., 2009). Finally, shoulder, neck and facial muscles should be relaxed once reached the upright position and full speed (Jones, et al., 2009).

EMG activity

Running requires a complex sequencing of body muscle activation. EMG activity has generally been found to increase with increased running speed (Mero, Komi, & Gregor, 1992). In the propulsion phase, EMG activity is markedly lower than during the braking phase. This may be partly related to increase recoil of elastic energy during the propulsion phase (Cavagna, et al., 1971). An example of muscle activation pattern during maximal speed running is presented in Fig. 8. There is high activity in the leg musculature before contact (Mero & Komi, 1987), as at the beginning of the contact phase, large impact forces occur. It is important, therefore, that the leg extensors muscles are highly activated and stiff prior to as well as at the moment of impact. Energy is transferred by the elastic elements of the musculo-tendinous complex from the braking to the propulsion phase, and utilisation of such elastic properties has been shown to be important in increasing explosive force production during contact (Cavagna, et al., 1971; Hunter, et al., 2005).

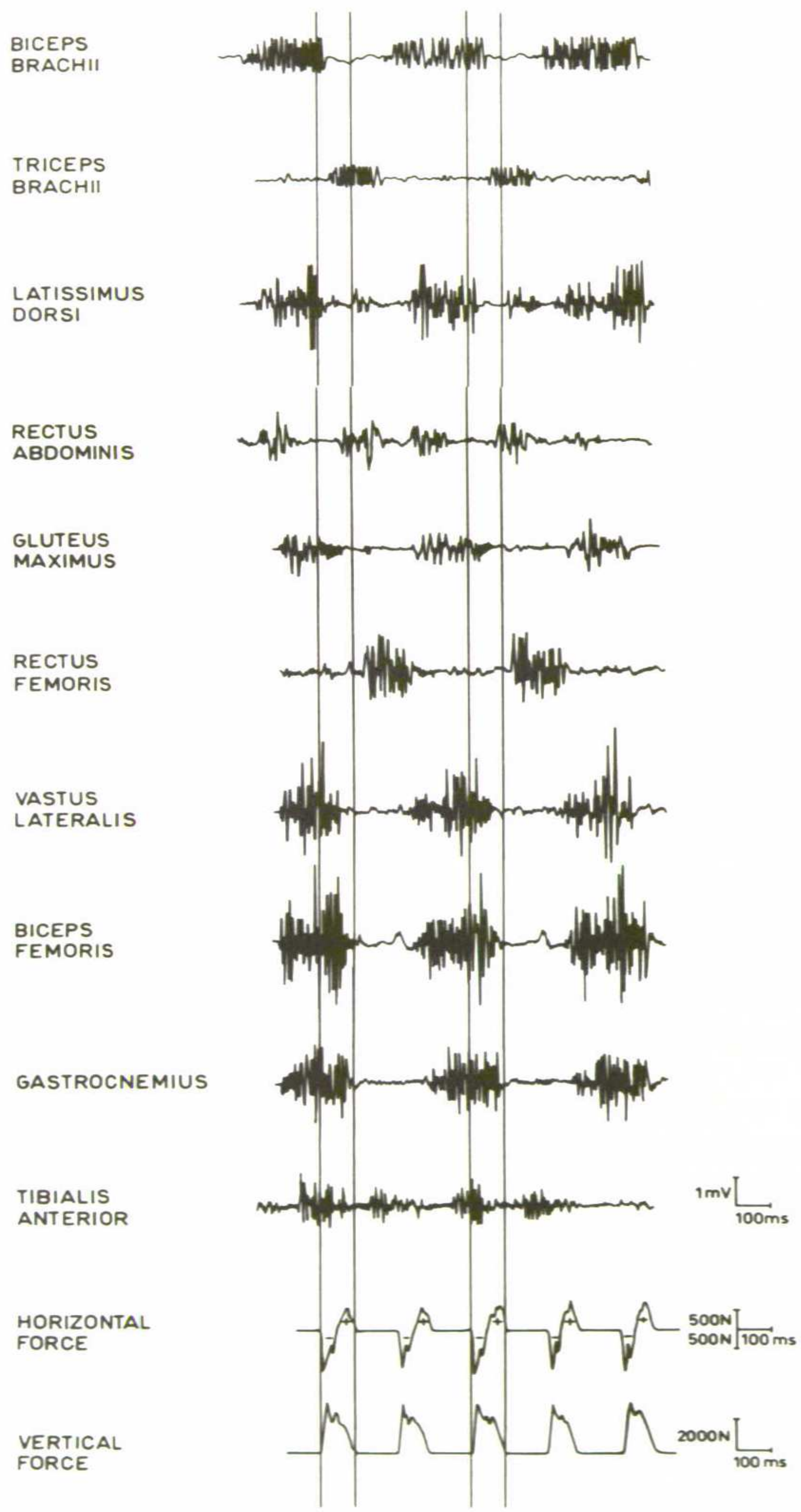


Figure 8 (1.2): Raw electromyographs of selected muscles during stride at maximal speed of one subject. Ground reaction forces measured by a force platform are also reported (from Mero, Komi, & Gregor, 1992).

Peak activity of the leg extensor muscles occurs during the braking phase of ipsilateral contact (leg on the ground). Thereafter, integrated EMG begins to decrease towards the end of the propulsion phase. The rectus femoris muscle seems to be more important as hip flexor than as knee extensor (Mero & Komi, 1987). Finally, the biceps femoris and gastrocnemius muscles are fairly active during the ipsilateral propulsion phase, and seem to play a primary role in the propulsion phase itself (Mero & Komi, 1987).

Kinematics and kinetics

Due to the difficulty of accurately reproduce a real sprint run in a laboratory environment, there are still many open issues about the kinematics and kinetics of sprint running during the maintenance phase. Nevertheless, a number of biomechanical variables have been identified as correlated with this phase performance.

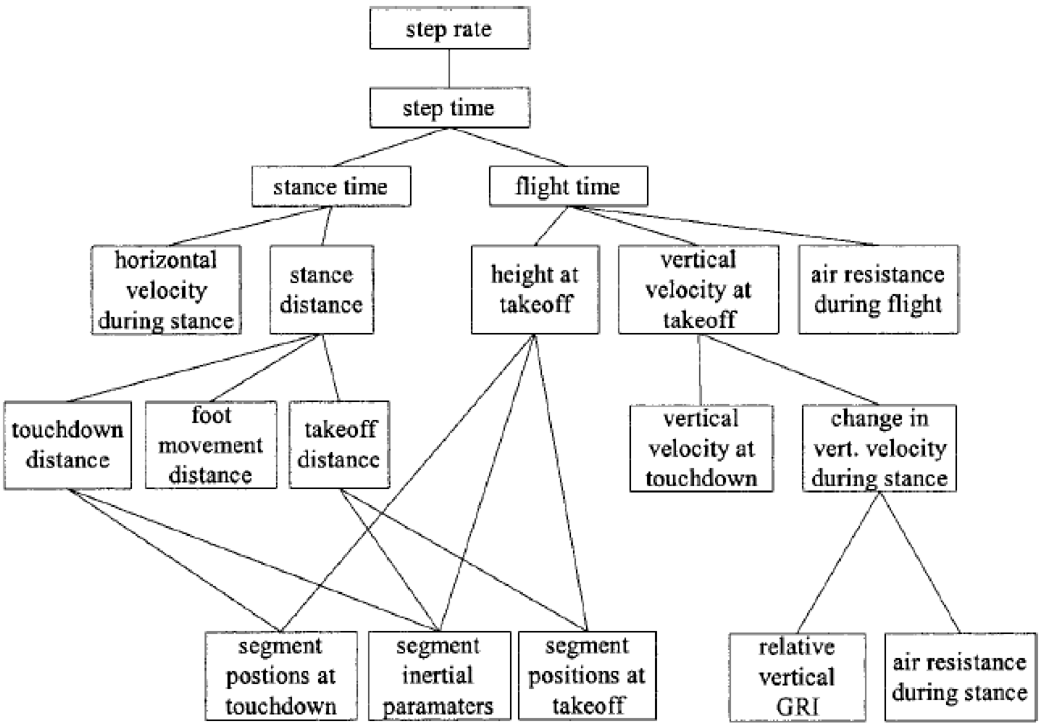


Figure 9 (1.2): Determinants of step rate (GRI: Ground reaction force Impulse) (from Hay, 1993).

- Step rate and step length: running velocity is the product of step rate and step length. Fig. 9 and Fig. 10 show the determinants of both parameters and how those determinants influence each others. In studies where the same subject ran at different speeds, both step rate and step length increased with speed (Luthanen, & Komi, 1978). This increase is linear for speeds up to 7 ms^{-1}. At higher speeds there is a smaller increment in step length and a greater increment in step rate for a given increase in velocity. This means that high speed runners tend to increase their velocity by augmenting step rate to a relatively greater extent than step length. At maximal velocities, in fact, it is suggested that step rate has a more decisive role than step length (Mero, Komi, & Gregor, 1992).

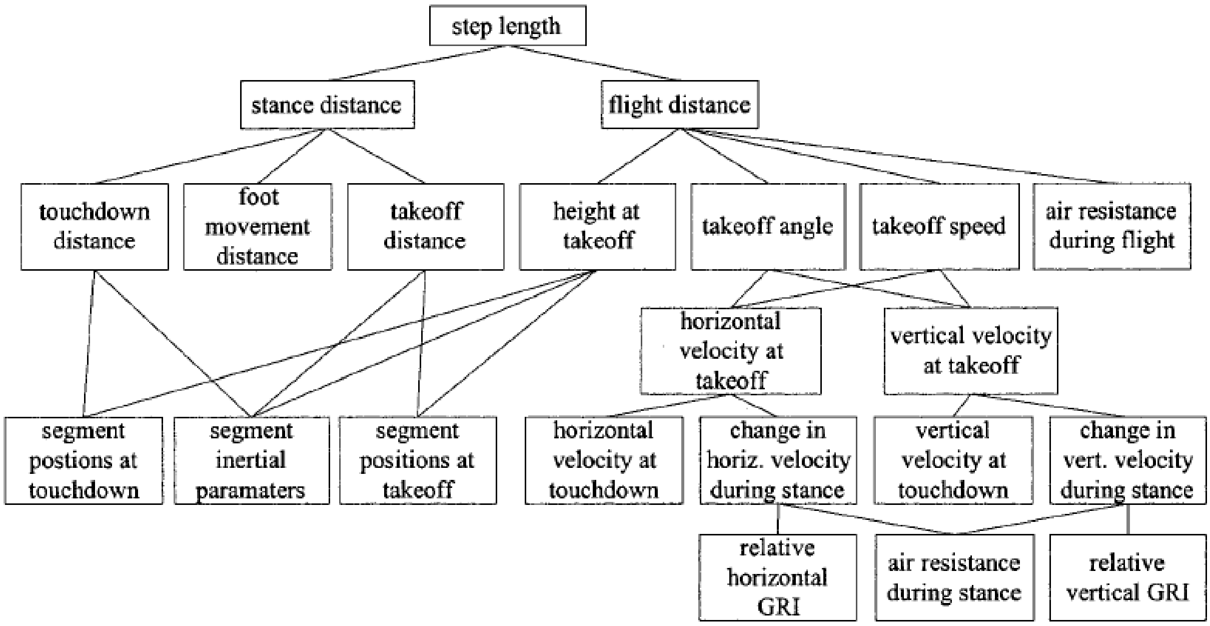


Figure 10 (1.2): Determinants of step length (from Hay, 1993).

Hunter et al. (2004b) investigated the relative influence of step length and step rate on race performance during the acceleration phase and determined the sources of negative interaction between these two parameters. Leg length, CoM height of takeoff, and CoM vertical velocity of takeoff proved to be all possible sources of a negative interaction between step length and step rate. The very high step lengths and step rates achieved by elite sprinters appeared to be possible only by a technique that involved high horizontal and low vertical velocity of takeoff. However, a greater vertical velocity of takeoff might be of advantage when an athlete is fatigued and struggling to maintain a high step rate. Mean step rate and step length values showed to be 4.26 samples per second and 1.91 m respectively (Hunter, et al., 2004b).

- Stance and flight times: stance time has been reported to decrease significantly as running velocity increases (Luthanen, & Komi, 1978), while flight time has the opposite behaviour. In the maintenance phase flight time ranges from approximately 0.120 to 0.140 s (Moravec et al., 1988; Hobara et al., 2009). In Fig. 11, stance and flight times of an elite sprinter performing a 100 m run are plotted against time. The relevant decreasing and increasing pattern of the two parameters is visible. Times were obtained by using a high-speed camera (Bergamini E., unpublished personal observations).

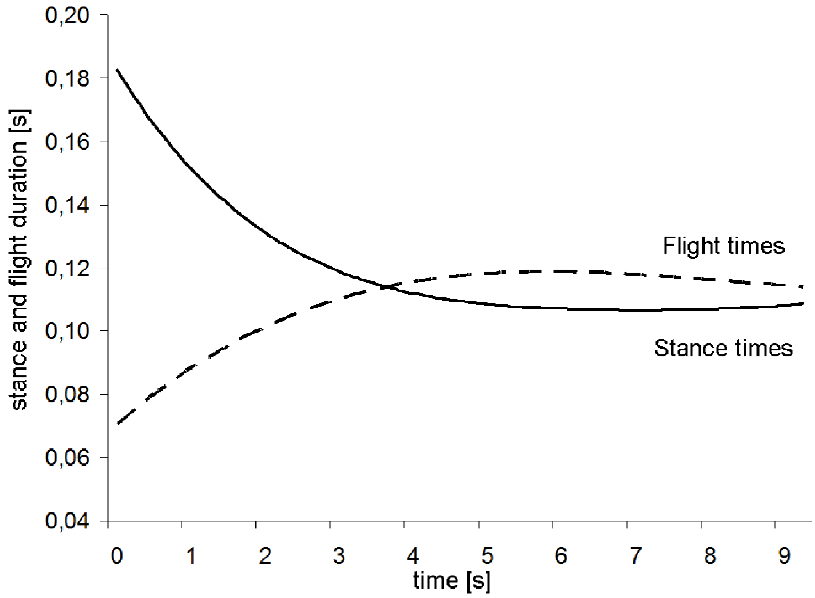


Figure 11 (1.2): Stance (solid line) and flight (dashed line) duration time-curve during a 100 m sprint run of an elite sprinter (Bergamini E., unpublished personal observations).

- Vertical displacement of the CoM: vertical peak-to-peak displacement of the CoM within the step duration has been shown to decrease with increased running speed (Cavagna, et al., 1971; Luthanen, & Komi, 1978). Mero et al. (1992) observed vertical displacements of 0.047, 0.050 and 0.062 m for “good”, “average” and “low level” male sprinters, respectively. Similar results were found by Hobara et al. (2009).
- CoM velocity: a number of studies found that at constant speed there is a decrease in the CoM horizontal velocity following initial foot contact. Then,

during the propulsion phase, the CoM velocity increases. Cavenagh and Lafortune (1980) found a decrease of 0.18 m·s^{-1} in running velocity during the braking phase at 4.47 ms^{-1}, which was followed by an increase of 0.27 m·s^{-1} during the propulsion phase. Velocity at toe-off was greater than that at touch-down. Probably due to the air resistance, during the flight phase the horizontal velocity of the CoM was found to decrease.

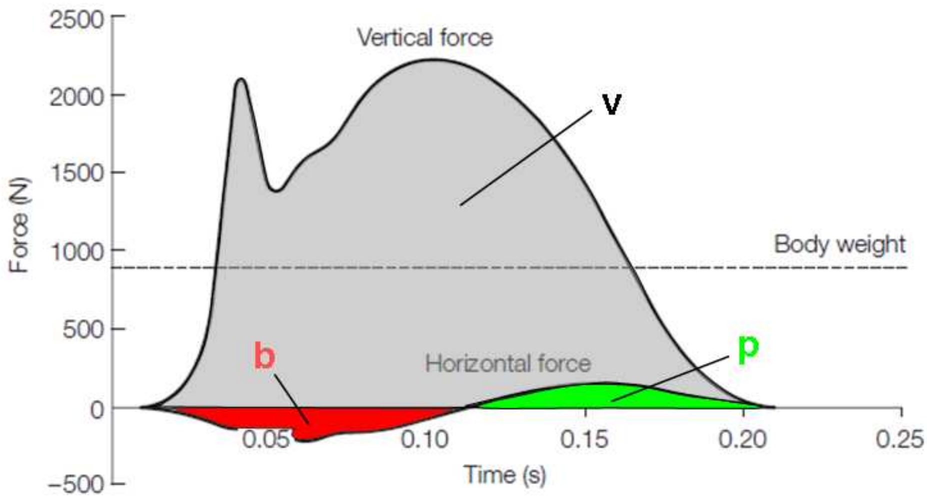


Figure 12 (1.2): GRF impulses are shown as areas under the horizontal and vertical GRF curves. p (green) is the propulsive impulse, b (red) is the braking impulse. Propulsive impulse was based on horizontal positive force data during stance, and braking impulse was based on horizontal negative force data during stance. Horizontal impulse was calculated as propulsive impulse minus the absolute value of braking impulse. v (grey) is the area under the vertical GRF curve. When horizontal, braking, propulsive, and vertical impulses are expressed relative to body mass, they reflect the change in velocity of the center of mass (ignoring the effects of wind resistance) during the respective periods and in the respective directions (adapted from Zernicke & Whiting, 2000, and from Hunter, et al., 2005).

- Force production: as during the previously analysed phases of the sprint run, force production has a key role during the maintenance phase as well. It has been reported that faster top running speeds are achieved with greater GRFs, rather than with rapid leg movements (Weyand, Sternlight, Bellizzi, & Wright, 2000). However, it has also been reported that large GRFs as well as high vertical and horizontal propulsive impulses are involved in stress fractures and knee-related injuries (Ferber, Kingma, Bruijn, & van Dieen, 2002; Grimston, Nigg, Fisher, & Ajemian, 1994). The ability of the sprinter to reduce the braking horizontal forces and to maximise the propulsion forces is crucial for the race performance (Hunter, et al., 2005). The braking horizontal force and braking time should be very small to avoid loss of velocity during the impact phase. Significant correlations were found between the average net resultant force in the propulsion phase and velocity, as well as between the propulsive force and the step length. Fig. 12 shows typical horizontal and vertical GRF curves, with relevant force impulses, for a well-trained sprinter (Hunter, et al., 2005).

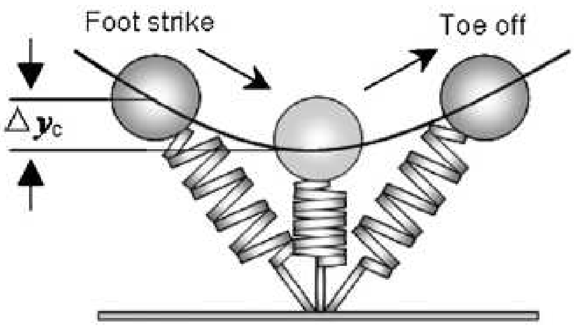


Figure 13 (1.2): Spring-mass model using in running to estimate vertical stiffness. The leg spring is compressed during the first half of the stance phase and rebounds during the second half. Maximal vertical displacement of the CoM during ground contact is represented by Δy_C (from Hobara, et al., 2010).

- Stiffness: The concept of stiffness has its origin in physics, as part of Hooke's Law. Objects that obey this law are deformable bodies which store and return elastic energy. Hooke's Law, defined as $F=k\cdot\Delta y$, states that the force (F) required to deform a material is related to a proportionality constant (k) and the distance (y) the material is deformed, provided that its shape is not permanently changed. The proportionality constant, k, is referred to as the spring constant, and it describes the stiffness of an ideal spring and mass system. The leg is often modelled as a spring supporting the mass of the body (Fig. 13). Lower extremity stiffness was proved to be an important and sensitive factor in musculoskeletal performance (Latash & Zatsiorsky, 1993; Butler, et al., 2003; Brughelli & Cronin, 2008). It appears, indeed, that increased stiffness is beneficial to performance. At the same time, too little or too much stiffness may increase the risk of musculoskeletal injury (Butler, et al., 2003). Although there is not a general consensus about the effect of increased or decreased stiffness on the risk of injuries, there is some evidence that increased stiffness may be related to bony injuries and decreased stiffness may be associated with soft tissue injuries (Burr, Martin, Schaffler, & Radin, 1985; Grimston, Nigg, Fisher, & Ajemian, 1994; Williams, McClay, & Hamill, 2001; Williams, Davis, Scholz, Hamill, & Buchanan, 2004). It appears to exists, however, an optimal amount of stiffness that allows for injury-free performance (Butler, et al., 2003). Moreover, it has been suggested that stiffness can be modified in response to the external environment or verbal cues (Hewett, Stroupe, Nance, & Noyes, 1996; Smith, & Watanatada, 2002).

1.2.4 Deceleration phase

Concerning the deceleration phase technique, no particular or specific aspects have been underlined in the literature. Athletes' aim and instruction is to limit as much as possible the speed decrease. This is particularly true in the 200 and 400 m races, where the athlete's endurance characteristics and anaerobic metabolism are crucial.

Very few studies focused on the deceleration phase of sprint running from both a physiological and biomechanical point of view. Recent progress as concerns technology and methodology will hopefully contribute in enriching the literature about physiological and biomechanical variables which influence and determine this phase performance.

EMG activity

EMG activity pattern during the deceleration phase was found to be similar with respect to that in the maintenance phase (Mero & Peltola, 1989). The average maximal activity of the muscles during the deceleration phase decreased of 6.8% with respect to that estimated during the maintenance phase of a simulated 100 m race.

Kinematics and kinetics

The deceleration phase is characterised by a decrease of velocity and by a slight increase in the stance times, and, in turn, a slight decrease in the flight times. In short sprint running events (especially in the 100 m race), the deceleration phase is usually visible on the time-velocity curve. The loss of velocity from the peak during a 100 m race ranges from 0.9 to 7% (Moravec et al., 1988). Step rate decreases, and step length slightly increases during deceleration (Moravec et al., 1988; Mero & Peltola, 1989). The CoM vertical displacement was also found to increase during this phase (Moravec et al., 1988; Mero & Peltola, 1989). No research data are available concerning the force production and the GRF patterns during the deceleration phase.

1.3 METHODS FOR SPRINT RUNNING ANALYSIS

This section was written on the basis of a recent review article by Higginson, 2009, on methods currently used for running gait evaluation. The author's work is gratefully acknowledged.

The remarkable increase in the number of recreational and competitive runners in recent years has obvious implications to professionals such as clinicians, physical therapists, and coaches who offer services aimed at the evaluation and rehabilitation of running-related injuries and performance enhancement strategies. Recent technological advancements now facilitate a greater range of individuals able to provide these services. With this increased demand comes an increase in the need to be knowledgeable of the technology currently available to provide these services.

Many tools have been developed to assist in the assessment of running gait. These include the more traditional motion capture systems used to describe motion of the body, force plates that quantify the forces acting on the body, and electromyography (EMG) used to estimate the level of muscle activity during motion.

More recently, smaller portable sensors have been developed and successfully used to measure running gait parameters. These include accelerometers, electrogoniometers, gyroscopes, and in-sole pressure sensors. These tools have been successfully used to investigate shoe (Verdejo & Mills, 2004; Butler, Davis, & Hamill, 2006; Clinghan, Arnold, Drew, Cochrane, & Abboud, 2008;

Dixon, 2008) and orthotic (Mündermann, Nigg, Humble, & Stefanyshyn, 2003) performance, risk factors for injury (Milner, Davis, & Hamill, 2006), running performance (Stafilidis & Arampatzis, 2007), fatigue effects (Derrick, Dereu, & McLean, 2002; Le Bris et al., 2006), and gait adaptations to various running techniques (Karamanidis, Arampatzis, & Brüggemann, 2004). For a summary of the specific gait parameters measured by each system and sensor, refer to the Tab. 3.

Sensor/System	Measured Parameters
Motion analysis systems	Segment position and orientation, linear and angular velocity, and acceleration
Force platforms	Ground reaction force, loading rates, center of pressure, joint moment, and power (when used with segment position and orientation data)
Pressure sensors	Pressure distribution, vertical force, center of pressure, spatiotemporal gait parameters
Electromyography	Muscle activation and timing patterns, muscle fatigue
Accelerometers	Segment acceleration and orientation, spatiotemporal gait parameters
Electrogoniometers	Relative joint angles
Gyroscopes	Segment orientation, angular velocity, and acceleration

Table 3 (1.3): Summary of running gait parameters measured by currently used systems or sensors (from Higginson, 2009).

In the next section a brief overview of the current technologies used in the assessment of running gait will be provided, with a focus upon the latest developments and equipment.

1.3.1 Electromyography

Electromyography (EMG) is a technique commonly used to measure levels of muscle activity during walking or running gait. Typically, timing of muscle activation and relative intensity are the primary measures of interest and can be collected through the use of surface or indwelling (fine-wire) electrodes. This technique can be used to detect abnormal gait behavior and assess the neuromuscular control of a runner (Dugan & Bhat, 2005).

Normal muscle activation during the stride and stance phases of running (Cavenagh, 1990; Novacheck, 1998) and sprinting (Mero, et al., 1992) have been reported (see also § 1.2), as well as how muscle activation and timing patterns are influenced by changes in walking and running speeds (Cappellini, Ivanenko, Poppele, & Lacquaniti, 2006). In addition to providing information about muscle activation levels and timing, the frequency content of the EMG signal can be analyzed to determine relative muscle fatigue (Wakeling, Pascual, Nigg, & von Tscharner, 2001), which may be used in the early detection of potential running injuries. There are also data to suggest that EMG parameters may be a more sensitive measure than force measures in explaining differences in shoe/orthotic comfort ratings (Mündermann, et al., 2003).

EMG parameters have been found to be highly reproducible between step cycles when compared across different running techniques (running velocity and stride frequency), but tend to be less reproducible when compared across individual muscles, with distal leg muscles showing more reproducibility than proximal leg muscles during running (Karamanidis, et al., 2004). Depending upon the particular muscle of interest, this information may influence selection of gait analysis protocols by decreasing (or increasing) the number of stride cycles needing to be analyzed.

One major drawback of older EMG systems is that data were transmitted via cable, potentially limiting the motion of the subject. Newer systems incorporate technology that allow for the data to be transmitted wirelessly or stored in a data logger worn by the subject, vastly increasing the functionality of these systems. Other limitations include crosstalk between muscles and electrical noise from external sources (Harris, & Wertsch, 1994). The incorporation of in-line preamplification devices has greatly reduced ambient noise in the underlying myographic signal, allowing for greater signal-to-noise ratio. Although selection of appropriate signal processing and normalization techniques, selection of muscle onset-offset determination algorithms, and data interpretation are a quite intricate task to inexperienced operators, many current systems are designed to accommodate various levels of user proficiency.

1.3.2 Motion analysis (Stereophotogrammetry)

The most common method for collecting information about position and orientation of body segments in two- or three-dimensional space is the use of motion capture technology (often referred to as stereophotogrammetry), in which markers are affixed to the subject and tracked throughout the motion of interest. These systems typically use passive markers that reflect ambient or infrared light. Through manual or automatic digitization techniques, the coordinate location (two- or three-dimensional) of the markers can be determined. From these position data, the velocity and acceleration can then be calculated by taking the time derivative of the position and velocity, respectively.

Limitations inherent in both optical systems include the need of skilled operators and the fact that they can be prohibitively expensive, have a relatively small capture volume, and require a controlled environment in which to operate (Sabatini, Martelloni, Scapellato, & Cavallo, 2005). Moreover, even when the motor task of interest can be performed and analysed in a human movement laboratory, sports motor acts are particularly problematic, being characterized by high accelerations and forces. The movement of the soft tissues, and in turn of the skin-mounted markers, relative to the underlying bones (soft tissue artefact - STA) appears to be one of the major sources of error in the estimation of kinematic and, particularly, kinetic parameters in the analysis of sports techniques (de Leva & Cappozzo, 2006). In this respect, there is a need for research in the development of new methodologies specifically designed for sports applications.

Edge detection technology recently has shown promise in allowing gait parameters to be quantified directly from video sources alone, allowing the same information to be extracted as that from marker-based systems, without the associated limitations or need for markers.

As concerns the use of treadmills, although they are often used in the analysis of walking and running gait to overcome issues surrounding small capture volumes, their use is believed to induce gait adaptations, such as increased time in stance phase, that normally would not be observed in over-ground running (Dugan & Bhat, 2005). These changes appear to be speed dependent (Nigg, De Boer, & Fisher, 1995). Moreover, sprint running can be hardly reproduced on treadmills.

To overcome problems related to the small capture volume, the use of new small and portable technologies seems to be effectively ideal.

1.3.3 Force plates

Force plates are commonly used to measure contact forces between foot and ground (ground reaction force – GRF). This information can be used to quantify impact forces, loading rates, as well as propulsive and breaking forces, and to track changes in the center of pressure (CoP) over time.

Because of their relatively small size, however, they impose constraints on foot placement, which may result in subjects adopting a "targeting" strategy while running, altering natural gait mechanics (Paolini, Della Croce, Riley, Newton, & Casey Kerrigan, 2007). This targeting strategy can lead to increased step length variability (Wearing, Urry, & Smeathers, 2000) and often results in exclusion of the trial (Milner, Ferber, Pollard, Hamill, & Davis, 2006), resulting in prolonged data collection periods and subject fatigue. Although the influence of targeting shows little effect upon GRFs during walking gait (Wearing, et al., 2000), greater approach velocities associated with running and larger changes in relative stride length to contact the force platform may result in more significant differences observed in GRFs at these higher speeds.

Recently, the development of instrumented treadmills have allowed for the quick collection of GRFs over repeated gait cycles, allowing for highly controlled gait speed, while eliminating potential error introduced by targeting strategies. One of the major drawbacks of these systems, however, is that treadmills using force plates beneath the belt can be susceptible to noise because of friction of the belt moving over the plate. Moreover, as above mentioned, sprint running can be hardly reproduced on treadmills.

1.3.4 Pressure sensors

The use of in-shoe pressure sensors provides a lightweight, portable, and easy-to-use alternative in which to analyze running gait. Unlike force platforms, they are capable of quantifying the distribution of force over the plantar surface of the foot, providing more detailed information on the loading of the foot during gait than force measures alone. Because this device is placed in the shoe, the loads acting on the foot surface can be measured directly, as opposed to the force acting on the bottom of the shoe with a standard force platform (Dixon, 2008). For this reason, in-shoe pressure sensors are commonly used to quantify the effect of shoe design upon foot loading and have been used to compare loading of the foot between similar shoe types (Clinghan, et al., 2008) and between shoes of different midsole designs (Dixon, 2008), as well as changes in the impact absorbing capabilities of a shoe over repeated impact cycles (Verdejo & Mills, 2004).

In-shoe pressure sensors also provide the ability to measure vertical forces experienced by the foot during prolonged running (Karamanidis, et al., 2004) and detect typical gait parameters required for gait analysis, such as heel strike and toe-off needed to define the stance phase of gait (Catalfamo, Moser, Ghoussayni, & Ewins, 2008). Although force platforms are considered to be the gold standard method by which these measurements typically are collected, as mentioned previously, they are limited in the number of steps that can be sampled and typically their use is restricted to a laboratory setting. In-shoe pressure sensors give the researcher or clinician the flexibility to collect data from repeated foot strikes in an environment that facilitates normal running gait.

Plantar loading parameters obtained from in-shoe pressure sensors have shown high reliability across multiple trials of the same subject, with low variability between steps (Murphy, Beynnon, Michelson, & Vacek, 2005), and are repeatable between testing days (Putti, Arnold, Cochrane, & Abboud, 2007). Analysis of GRF parameters indicates that these measures are reliable, over a range of running speeds and stride frequencies (Karamanidis, et al., 2004), also when collected using pressure sensors. However, comparison of the two most popular in-shoe pressure measurement systems indicates that the accuracy and precision of these systems may be sensitive to the levels of pressure applied, calibration procedure, duration of pressure application, and insole age of use (Hsiao, Guan, & Weatherly, 2002).

1.3.5 Accelerometers

The use of body-fixed sensors such as accelerometers are rapidly becoming a viable alternative to more traditional gait analysis techniques for use in the assessment of human motion. Accelerometers are inertial sensors that provide a direct measurement of acceleration along single or multiple axes, effectively reducing the error associated with differentiation of displacement and velocity data derived from sources such as motion capture systems.

Accelerometer-based systems have been used successfully to quantify the shock experienced by the lower extremity during walking and running (Lafortune, Henning, & Valiant, 1995; Butler, et al., 2006; Milner, Davis, et al., 2006), evaluate the effect of footwear (Butler, et al., 2006) and insoles (O'Leary, Vorpahl, & Heiderscheit, 2008) upon tibial shock during running, evaluate shock attenuation between body segments during running (Mercer, Bates, Dufek, & Hreljac, 2003), and investigate the effects of fatigue upon running gait patterns (Le Bris, et al., 2006).

Perhaps the most appealing advantage is the ability of accelerometers to be used in the estimation of spatio-temporal gait parameters (Sabatini, et al., 2005), which until recently required the use of a force plate, motion analysis systems, or footswitches. As addressed in previous sections, a primary limitation of stereophotogrammetric systems and force plates in the analysis of running gait is their limited ability to measure successive strides. Because of their light weight and portability, accelerometers are capable of recording data that can be collected continuously over many stride cycles for a prolonged period of time. This technology has been used effectively to detect alterations in running patterns after the onset of fatigue in middle-distance runners while running on a track, without altering the running patterns of the runner (Le Bris, et al., 2006).

Although mechanical testing has confirmed the validity and reliability of accelerometers in the measurement of accelerations within the frequency and amplitude range of human body motion (Bouten, Koekkoek, Verduin, Kodde, & Janssen, 1997), evidence indicates that they are sensitive to the site and method of attachment, with skin-mounted accelerometers resulting in significantly greater peak accelerations than bone mounted accelerometers (Lafortune, 1991). Moreover, direct comparison of axial acceleration of the tibia during running found that the disparity in results between attachment methods is largely subject-dependent (Lafortune, et al., 1995). Another drawback of these sensors is that the acceleration signal is affected by a fluctuating offset (bias) and random white noise, which may jeopardise the result of the numerical integration process necessary when velocity or displacement has to be estimated (Woodman, 2007).

1.3.6 Gyroscopes

Gyroscopes are miniature angular rate sensors that can be attached to individual body segments, providing a direct measure of segment angular velocity. This sensing technology is an inexpensive alternative to motion analysis systems (Mayagoitia, Nene, & Veltink, 2002), and methods recently have been developed to calculate spatio-temporal gait parameters based upon the angular velocity measures provided by these sensors (Aminian, Najafi, Büla, Leyvraz, & Robert, 2002; Sabatini, et al., 2005). Little measurement error is entailed in estimating these parameters, in comparison with measurements based on foot pressure sensors (Aminian, et al., 2002).

In addition to allowing the direct measurement of segmental angular velocity and calculation of spatio-temporal gait parameters, other benefits of using gyroscopes include their small size and portability, lower power requirement (Wong, Wong, & Lo, 2007), and insensitivity to gravitation influence (Sabatini, et al., 2005). However, as with accelerometers, signal integrity is compromised by unwanted sensor motion resulting from poor fixation or site selection and from the bias and random white noise. When integrating the angular velocity signal in order to estimate the angular displacement, therefore, drift errors can be crucial.

Gyroscopes and accelerometers utility can be further enhanced through the concurrent use of the two sensors. Sensor fusion theory, borrowed from the aerospace field, has been applied for estimating sensor orientation, with accelerometers used for compensating the drift which affects the angular displacement obtained by numerical integration of noisy gyroscope signals, by using gravity as an absolute reference direction. The latter assembly is commonly referred to as inertial measurement units (IMU) (further details are reported in § 3.1.2). When combined with rate gyroscopes, accelerometers have been found to produce joint angle, angular velocity, and angular acceleration similar to that derived from motion capture systems under dynamic conditions (Mayagoitia, et al., 2002), and have been used effectively to estimate walking speed and surface inclination angles (Sabatini, et al., 2005).

1.3.7 Electrogoniometers

Electrogoniometers allow for the direct measurement of joint angles during continuous dynamic activities. They offer a simple, affordable alternative to motion capture systems and allow joint angle data to be collected and viewed instantaneously. The end blocks of the electrogoniometers typically are affixed to the skin on either side of the joint axis of rotation using double-sided adhesive tape, as specified by the manufacturer. This method has been found to result in excessive sensor motion (Rowe, Myles, Hillman, & Hazelwood, 2001), which potentially could be magnified during the high-speed changes in joint angle commonly experienced during running. This unwanted sensor motion can be effectively reduced through the use of additional adhesive tape (Piriyaprasarth, Morris, Winter, & Bialocerkowski, 2008) or application of pre-wrap and athletic tape (Dierick, Penta, Renaut, & Detrembleur, 2004). For prolonged periods of data collection, special suits have been fabricated, which facilitates attachment of the electrogoniometers using hook and loop fasteners (Pierre et al., 2006). Failure to prevent sensor motion and improper alignment of the sensor during the application procedure has been found to be the greatest potential contributor to measurement error (Rowe, Myles, Hillman, & Hazelwood, 2001; Wong, et al., 2007; Piriyaprasarth, et al., 2008).

Measurement error from electrogoniometers has been shown to be as small as 0.04 degrees (Piriyaprasarth, et al., 2008) and has been validated using both human (Rowe, Myles, Hillman, & Hazelwood, 2001) and mechanical (Piriyaprasarth, et al., 2008) testing protocols, with results comparable to those obtained using motion capture systems (Rowe, Myles, Hillman, & Hazelwood, 2001). Although studies have shown inter- and intra-tester reliability to be relatively high, it has been suggested that the same tester be used when possible to ensure the highest repeatability (Piriyaprasarth, et al., 2008). When applied correctly, electrogoniometers have proven to be highly accurate and highly sensitive for detecting changes in joint angles over time, providing a simple, small, portable, and affordable alternative to motion capture systems (Rowe, Myles, Hillman, & Hazelwood, 2001).

1.4 THEORETICAL BACKGROUND: DISCUSSION

In the last decades, due to the increase in running popularity, biomechanical variables influencing sprint running performance have been widely studied. In particular, the block start phase together with the first steps of the acceleration phase have been performed and analysed both in traditional human movement analysis laboratory and on instrumented tracks. The evaluation of the maintenance phase, however, requires the use of treadmills, thus inducing running technique adaptations that would not be observed in in-field running. This is particularly true at high speeds.

The lack of portability and the small acquisition volume characterising traditional laboratory-based instrumentations used for the analysis of human movement, therefore, still represent the main limitation to in-field evaluation of sprint running. Thanks to recent technological advances accelerometers, electrogoniometers, gyroscopes, inertial measurement units and in-shoe pressure sensors have been made available. Combined with wireless technology and/or data loggers, they provide an affordable, lightweight alternative to in-field running analysis, allowing data collection over prolonged periods of time in almost any environment. Investing resources in the development and validation of methods and protocols that exploit the use of such technologies, appears, therefore, fully appropriate.

On the other hand, only very few studies focuses on determining definitive relationships between either anatomical factors and injury, or biomechanical measures and injury in sprint running. The main result concerns the relationship between the amount of forces applied to the lower extremities and the incidence of knee injuries. Being the forces and accelerations involved in sports motor acts considerably higher with respect to those implicated in clinical contexts, the development of methods specifically designed for the estimation of internal and external forces in sports applications is, therefore, desirable.

CHAPTER 2

AIM OF THE THESIS

The present work is part of a co-tutoring PhD program developed by the University of Bologna, the Locomotor Apparatus Bioengineering LABoratory (LabLAB) of the University of Rome "Foro Italico" and the Laboratoire de Biomécanique (LBM) of the Arts et Métiers ParisTech of Paris, France.

The general purpose of the present thesis is to contribute overcoming the main limitations concerning the biomechanical analysis of sports motor tasks focusing, in particular, on sprint running evaluation. To this aim, two complementary approaches have been adopted:

- <u>LOW RESOLUTION APPROACH</u>: the necessity of performing in-field athlete evaluation without influencing or constraining athletes' activities will be coped by exploring the use of wearable inertial measurement units during sprint running. Relevant biomechanical variables affecting sprint running performance will be estimated.

- <u>HIGH RESOLUTION APPROACH</u>: the lack of methods specifically designed for sports applications will be coped by focusing on constraint-based knee joint models commonly used to enhance the accuracy of joint kinetic estimates, being the latter particularly crucial in sports injury prevention. To further improve such models, subject-specific non-rigid constraints for the knee joint will be defined.

***Note**: The term "Resolution" refers to the level of detail relative to the specific framework of analysis. Hence, the Low Resolution Approach entails a global assessment of the performance by using inertial sensors and a less detailed analysis is necessary. Conversely, the High Resolution Approach, focusing on the internal forces and joint kinematics, entails a more detailed observation.*

CHAPTER 3

LOW RESOLUTION APPROACH

- *V. Camomilla, G. Vannozzi, P. Picerno, E. Bergamini – "Wearable inertial sensors: sport biomechanics meets performance evaluation". In: Capranica, L. and Meeusen, R., Intensive Programme on Sport Performance: A Lifespan Challenge, eds. Routledge, Abingdom, UK.*

- *E. Bergamini, P. Picerno, V. Camomilla, H. Pillet, A. Cappozzo – "Estimate of temporal parameters during sprint running by trunk inertial sensing: in field validation", XXIX Conference of the International Society of Biomechanics in Sports, June 2011, Porto, Portugal.*
- *E. Bergamini, P. Guillon, H. Pillet, V. Camomilla, W. Skalli, A. Cappozzo – "The trunk orientation during sprint start estimated using a single inertial sensor", 28th Conference of the International Society of Biomechanics in Sports, July 2010, Marquette, Michigan (US).*
- *I. Masci, E. Bergamini, G. Vannozzi, N. Getchell, A. Cappozzo – "Quantitative assessment of locomotor skills development in childhood: the running paradigm", 7th Annual Biomechanics Research Symposium, May 2010, Newark, Delaware (US).*
- *E. Bergamini, P. Picerno, M. Grassi, V. Camomilla and A. Cappozzo – "Spatio-temporal parameters and instantaneous velocity of sprint running using a wearable inertial sensing unit", 27th Conference of the International Society of Biomechanics in Sports, August 2009, Limerick (Ireland).*
- *E. Bergamini, P. Picerno, M. Grassi, V. Camomilla and A. Cappozzo – "Estimate of performance correlated parameters in non steady-state sprint running using a wearable inertial measurement unit", 10th Congresso SIAMOC, October 2009, Alghero (Italy).*

ABSTRACT

Athlete's performance evaluation is one of the main issues of coaching, as well as of sports biomechanical analysis. An optimised performance is obtained when accurate, useful, and timely feedback about athlete proficiency is provided. To this aim, the main goal of sport biomechanists is to provide information to coaches and athletes about sports skill technique.

As the majority of sports motor acts are hardly repeatable in laboratory environments, in-field evaluation of the athlete performance, without influencing or constraining athletes' activities, is now becoming mandatory in order to provide coaches with reliable and realistic information. Sprint running is not an exception.

Thanks to their portability and weightlessness, inertial measurement units allow for data collection during unconstrained continuous movement over prolonged periods of time. Accelerometer and gyroscope signals measured by these sensors, however, are affected by two main sources of errors: an unstable drift that accumulates when numerical integration is performed and the effects of soft tissue wobbling relative to the underlying bones.

Three studies will be presented aiming at estimating reliable performance-related parameters by using inertial sensors and at developing methods able to reduce the effects of the above mentioned sources of errors. In particular, the trunk inclination and angular velocity, as well as the instantaneous horizontal velocity and displacement of a point approximating the center of mass will be provided during in-lab sprint running. The stride and stance durations will be estimated on-the-field during the maintenance phase.

The fixation of the sensor proved to be a crucial aspect in reducing soft tissue oscillations. The use of memory foam materials and of *ad-hoc* elastic belt appears effective. Reducing the integration interval and cyclically determining the initial conditions of the acceleration integration process proved to yield reliable instantaneous velocity as well as spatio-temporal parameters.

KEYWORDS: Performance evaluation; Sprint running; Sprint start; Inertial sensors; Spatio-temporal parameters.

3.1 INTRODUCTION

3.1.1 Sports biomechanics and in-field performance evaluation

Athlete's performance evaluation is one of the main issues of coaching. An optimised performance is obtained when accurate, useful, and timely feedback about athlete proficiency is provided. A successful coaching outcome is related to the ability of coaches and trainers to correctly analyse all the deterministic details of the performance. In this way, training programs can be organized to specifically target performance defects. Failure in providing such feedback implies a reduction of chances of improvement (Carling & Williams, 2005). Therefore, athlete monitoring, evaluation and training planning should be based on systematic, objective and reliable approaches to improve data acquisition and information processing. In this respect, a gap exists between sport science research and coaching practice (Goldsmith, 2000). The link between research and coaching practice needs to be reinforced, especially in élite sports, where coaches are progressively going to incorporate the outcomes of sports science research in their in-field activity (Williams & Kendall, 2007).

Sport coaches aim naturally at improving athlete's performance and at reducing injury risk. These two objectives are also shared by sport biomechanists. The science of biomechanics is concerned with the forces that act on human body and the effects these forces produce. Physical education teachers and coaches of athletic teams, whether they recognise it or not, are likewise concerned with causes (forces) and effects (movement). In fact, a sport technique can be defined as the way in which body segments move in relation to each other during a movement task. Coaches' ability to teach sport techniques depends largely upon their understanding of both the effects they are trying to produce and the forces that cause them. Thus, the main goal of sport biomechanists should be to provide information to coaches and athletes about sports skill technique that will assist them in obtaining a good technique, characterised by effective performance (the purpose of the movement) and decreased risk of injury (distribution of forces in muscles, bones, and joints so that no part is excessively overloaded). Poor techniques are characterised by increased risk of injury, even though performance may be effective, at least for a while.

Sport scientists basically use two approaches to analyse athletes' technique: qualitative and quantitative. A qualitative analysis is based on the systematic observation of the motor task, directly and/or via film or video (Knudson, 2007). The effectiveness of this approach depends on the operator ability to observe accurately the task and to know what to look for, i.e. on the operator knowledge and experience and, in particular, his/her ability to identify the mechanical requirements of the movement under analysis. Since human observation is generally not sufficient to provide accurate and objective information, the use of objective data, specific measuring tools, and correct interpretation and application of the findings are required to optimise the coaching process. This circumstance calls, therefore, for the use of quantitative analysis. A quantitative analysis is based on measurements of the kinematic (displacement, velocity, acceleration) and kinetic (force, moment, power) variables that determine performance. As regards the mechanical determinants of performance, relevant assessments have been, so far, typically constrained into laboratory settings. Laboratory-based instrumentations are accurate, but the volume of capture is limited and may even vitiate the execution of the motor task under analysis. Here is where sports biomechanics fails in providing useful and exploitable information to coaches and trainers.

A perfect example of the "breakpoint" between sports biomechanics and performance evaluation is represented by Mr. Pistorius with his emblematic 400 m race during the 2007 Golden League in Rome. After his brilliant vamp in the last 200 m, a group of researcher was requested (by the International Association of Athletics Federations, IAAF) to investigate about his sprint performance ability and determine whether or not he could have had mechanical advantages from his prostheses. Two groups of researchers performed different mechanical and physiological tests on Mr. Pistorius. The first expressed its positive opinion regarding the advantages coming from the prostheses, while the second group expressed the opposite attitude. This report was then followed by the sentence emitted by the Court of Arbitration for Sport (Lausanne, May 2008) which stated "*Having viewed the Rome Observations [...] the IAAF's officials must have known that*

[...] the results would create a distorted view of Mr. Pistorius' advantages and/or disadvantages by not considering the effect of the device on the performance of Mr. Pistorius over the entire race". In fact, the researchers' investigation was performed using lab-based instrumentations and, thus, was limited to few steps. Conversely, in cases like this, a mechanical characterisation of the motor task must be carried out for the entire race. The abovementioned circumstance might be overcome when adequate technological advancements are made available.

Technological advances and improvement of procedures have been and continue to be an important issue in sport and exercise science (Winter, et al., 2007). To cope with the fast development of new technological advancements, sports scientists and coaches are asked to continuously account for new developments and include them in the everyday assessment of training and competition. Specifically, technologies used to measure performance are moving forward and improved methods based on "state of the art" computer technology and robotic automation are being continually developed and commercialised to support the pursuit of success in elite sports (Carling, Reilly, & Williams, 2009). Systems of measurement benefitted of such trend with a progressive device miniaturisation, increased reliability and cost reduction. Simultaneously, data processing took advantages of greater calculation power, speed and interactivity. Last but not least, the development in the use of expert systems such as, for example, Artificial Neural Networks, as diagnostic tools for evaluating faults in sports techniques is advocated as promising in the near future (Bartlett, 2006).

Professionals from computer science and engineering fields cooperate with sports biomechanists and physiologists to define and implement performance indices that can be used by trainers and coaches during the performance evaluation exercise. From the same multidisciplinary interaction, further improvements were attained, such as instrumented sports equipments or portable systems that use telemetry for data transmission. These ones are particularly useful for in-field performance evaluation being not bound to laboratory conditions.

3.1.2 Wearable inertial sensors

An alternative to laboratory based techniques is the use of inertial sensors that can measure movement-related data without any space limitation and no cumbersome setup. Their measure is based on the reluctance (inertia) of a portion of the sensor to accelerate linearly (accelerometer, inertia to linear motion) or about an axis (gyroscope, inertia to angular motion) as a response to forces and torques impressed to them, respectively. The vestibular system, located in the inner ear, is a biological 3D inertial sensor. It can sense angular motion as well as linear acceleration of the head.

The measured quantities are expressed with respect to a moving reference frame defined by the sensitive axes of the sensors. The sensitive axis is the axis along which the acceleration is "sensed" and about which the rotational velocity is "sensed". Accelerometers and gyroscopes may become triaxial when three uniaxial sensors are mounted mutually orthogonal to each other. When a measure of a global (inertial) reference frame is required, the measure of the direction and of the intensity of a local magnetic field is also used.

Only recently advances in micro-electro-mechanical systems (MEMS) technologies have led to the development of a new generation of inertial sensors, the specifications of which - in terms of encumbrance, robustness, power consumption, measuring performance and cost - seem to be appropriate for applications in the biomedical field. Sensor fusion theory, borrowed from the aerospace field, was applied for estimating sensor orientation: accelerometers are used for compensating the drift which affects the angular displacement obtained by numerical integration of noisy gyroscope signals, using gravity as an absolute reference direction. The latter assembly is commonly referred to as inertial measurement units (IMU). Sensor fusion algorithms are then enhanced with the magnetic sense furnished by MEMS magnetometers, using the magnetic North as an absolute reference direction on the horizontal plane. With the latter assembly, also known as a magnetic field angular rate gravity (MARG) sensors (Bachmann, 1999), an absolute Earth-fixed reference frame is provided.

Currently, the major disadvantage of MEMS is the reduced performance in terms of accuracy and bias stability. Signals measured by MEMS sensors are, indeed, affected by a fluctuating offset (bias) and random white noise. Size and performance of an inertial sensor seem to be linearly dependent parameters, thus the smaller the sensor, the lower the performance expected. On the other side, MEMS technology transforms inertial sensors into body-fixed sensors. MEMS sensors are highly transportable and do not need any stationary units such as transmitters, receivers or cameras. All detected signals can be recorded by a portable data logger, allowing the subject to perform his/her activity in a real situation. There are no restrictions in the capturing volume and data can be recorded for long periods of time depending on the performance and capacity of the data logger. All these features allow mobile and outdoor motion capture. In addition, these sensors are much cheaper than sonic, magnetic and optical motion capture systems and present a high ecological validity: being worn by the subject, they can be used in non-standardized environments and on any surface (for e.g. grass, sand, balance beam, springboard, etc.). Moreover, little time is needed to set up the system and a large number of athletes can be evaluated in a short time in agreement with trainers' requirement.

For all these reasons, the availability of wearable motion sensors has opened new perspectives in sport sciences and, in particular, in the field of human movement analysis. In few years a large number of applications have appeared both in the literature and on the market. For example, vertical jump performance has been analysed using triaxial or uniaxial accelerometers either placed on the ankle (Quagliarella, Sasanelli, Belgiovine, Moretti, & Moretti, 2010) or on the sacrum (Innocenti, Facchielli, Torti, & Verza, 2006; Palma, Silva, Gamboa & Mil-Homens, 2008). However, all these assessments suffered from the limits of accelerometers and did not take advantage of the presence of gyroscopes that could greatly enhance the evaluation (Mazzà, Iosa, Picerno, & Cappozzo, 2009).

Strength and power measures can contribute in devising methods to improve power output and to support its transference to athletic performance (Cronin & Sleivert, 2005). Although it is still debatable whether training at maximal power improves functional performance, maximal power needs to be constantly monitored and adjusted. Isokinetic and isometric assessment bear little resemblance to the accelerative/decelerative motion implicit in limb movement during resistance training and sporting performance (Cronin & Sleivert, 2005), while measures performed using inertial sensors seem ideal to perform power assessment (Jidovtseff, et al., 2006).

Kinetic information derived from acceleration measures gain increased value when obtained directly on the field. A well fitting example is the system, constructed by Baca and colleagues, for recording and presenting relevant kinetic information during on-water and ergometer rowing (Baca, 2006). Similarly, IMUs have been used as training aid for hammer throw (Otha, Umegaki, Murofushi, Komine, & Miyaji, 2008), or, on the tennis court, as a tool to perform a motor strategy analysis, investigating the translational and rotational motion of the swing (Ahmadi, Rowlands, & James, 2006) or highlighting differences in the forehand stroke according to aim and direction (Camomilla, Lupi, & Picerno, 2008). Coupling accelerometers with other monitoring techniques could enhance their potential as a means to assess performance over the entire course. This approach was used, for example, by competition rowers to improve performance at national and international competitions, monitoring impeller velocity (James, Davey, & Rice, 2004). Moreover, foot-ground contact times (Auvinet, et al., 2002; Purcell, et al., 2005; Lee, Mellifont, & Burkett, 2010; Wixted, et al., 2010), lower leg rotational kinematics (Channells, et al., 2005), vertical stiffness (Hobara, et al., 2009) and running gait symmetry (Lee, Sutter, Askew, & Burkett, 2010) have been assessed using accelerometers on the tibia or on the lower-back trunk during distance and sprint running.

It must be kept in mind, however, that the extraction of movement-related information content from the signal derived from MEMS sensor can be strongly jeopardized by the unstable bias that characterises the signal of such sensors and by soft tissue vibrations. In the case of sport performance assessment, during the execution of explosive sport manoeuvres, the former plays a secondary role compared to the devastating effects of soft tissue vibration. For this reason, prior to perform any measurement, good practice rules recommend to mitigate this source of error for the benefit of the further signal processing. Developing algorithms that can minimise the detrimental effects of these errors, while transforming this data into well-known parameters suitable for use by coaches and athletes, remains the main challenge. Novel derivative techniques, pattern recognition of specific activities, and characterisation of specific phases of the task can greatly improve sensor data interpretation. This data reduction can lead to valuable information that can be made available to coaches and trainers directly on the field and during real scenarios, such as training sessions or race simulations.

3.2 <u>STUDY 1</u>: TRUNK INCLINATION DURING THE SPRINT START USING AN INERTIAL MEASUREMENT UNIT

Abstract

The execution of the start is crucial in determining the performance during sprint running. In this respect, trunk inclination is a key element in moving from the crouch to the upright position. The purpose of this study was to provide coaches with an instrument able to reliably estimate such parameter in-field. To this aim, the accuracy of an inertial measurement unit (IMU) in estimating its rotation about a local axis and the relationship between this rotation and trunk inclination in the progression plane were assessed during block start and pick-up phases.

Sprint starts were performed by five sprinters and data provided by an IMU positioned on the trunk at L2 level were compared to reference stereophotogrammetric measurements. An attachment that limited soft tissue oscillations and Kalman filtering were used to reduce errors. The trunk was modeled as a rigid segment joining the midpoint between C7 and the posterior superior iliac spines.

The accuracy of the IMU in estimating its orientation about a local axis was high both in terms of curve similarity (correlation r>0.99) and of bias (lower than 1 deg). Similar results were obtained concerning the relationship between the IMU estimates and the trunk inclination in the progression plane (r>0.99; bias lower than 4 deg). Results open a promising scenario for an accurate in-field use of IMUs for sprint start performance assessment.

Keywords: Inertial sensor; Kalman filter; Performance indicator; Sprint start; Biomechanics.

3.2.1 Introduction

The proper execution of the sprint start is crucial in determining the performance during the 100, 200 and 400 m track sprint runs (Harland & Steele, 1997; Ostarello, 2001; Čoh, et al., 2006). The main outcome of the sprint start is to move from a static crouched position to a cyclic upright one, by generating sufficient upward vertical velocity and, at the same time, by creating maximal forward horizontal velocity to improve the sprint performance. To reach high horizontal velocities, sprinters have to exert high horizontal forces through the blocks in a short time period (Harland & Steele, 1997; Čoh, et al., 1998; Kraan, van Veen, Snijders, & Storm, 2001).

Different parameters were obtained and were shown to be correlated with the athlete time at 20 m, divided in the block start phase, ranging from when the athlete obtains the "on your marks" position to block clearing, and pick-up phase, ranging from the block clearing to the instant of time in which the athlete attains an upright sprinting position:

- the relative position and the inclination of the blocks (Schot & Knutzen, 1992; Cousins & Dyson, 2004; Čoh, et al., 2006);
- the vertical and horizontal trajectory and velocity components of the athlete's centre of mass (Natta & Breniere, 1997; Cousins & Dyson, 2004; Čoh, et al., 2006; Kugler & Janshen, 2010);
- the force impulse exerted on the blocks during the block phase (Kugler & Janshen, 2010; Slawinski, Bonnefoy, & Levêque, 2010);
- the reaction time, i.e. the time from the gun signal to the first detectable change of pressure on instrumented blocks, and the block time, i.e. the total time the athlete spends on the blocks, from the first detectable change of pressure to the front block clearing (Mero, et al., 1992; Fortier, et al., 2005; Čoh, et al., 2006);
- the time between the onset of leg muscle activity and that of force production (Mero & Komi, 1990).

All these parameters, however, are related entirely or mainly to the lower part of the body. Only recently, Slawinski et al. (Slawinski, Bonnefoy, Ontanon, et al., 2010) reported that, since the upper body kinetic energy contributes significantly to the total body kinetic energy, the coordination between the upper and the lower part of the body is a key factor in the block start performance. From coaches' perspective, it is acknowledged that the upper part of the body is also important. In fact, expert track and field coaches focus primarily on the inclination of the trunk in the progression plane described as the line joining the shoulder and the hip joint centers (Jones, Bezodis, & Thompson, 2009).

Only a few quantitative studies have focused on the trunk during the sprint start (Mero, Luhtanen, & Komi, 1983; Čoh, et al., 1998; Natta, Decker, & Boisnoir, 2006; Slawinski, Bonnefoy, Ontanon, et al., 2010). Trunk inclination in the progression plane was analysed during the set position (Mero, Luhtanen, & Komi, 1983; Čoh, et al., 1998; Natta, Decker, & Boisnoir, 2006) and, on a sample of female athletes, was found to correlate with the athletes' time to 10 and 20 m from the starting line (Čoh, et al., 1998). Natta and colleagues (Natta, Decker, & Boisnoir, 2006) reported that the trunk angular velocity, analysed during the block start, discriminates between high and medium level sprinters. None of these studies monitored the trunk inclination and angular velocity during both block start and pick-up phase. Moreover, data acquisition was performed using motion capture or video analysis technologies, both limited in terms of acquisition volume, post-processing time, and cost.

As an alternative to video-based techniques, wearable inertial measurement units (IMUs) have recently gained momentum as a suitable solution for low cost in-the-field biomechanical analysis of running performance. They provide, without any space limitation and cumbersome setup, 3D linear acceleration and 3D angular velocity with respect to a local unit-embedded system of reference. So far, IMUs have been used to estimate foot-ground contact times (Purcell, et al., 2005) and lower leg rotational kinematics (Channells, et al., 2005) during sprint, and to estimate runners' speed (Vetter, Onillon, & Bertschi, 2009), symmetry between right and left lower limb (Lee, Sutter, Askew, & Burkett, 2010), and stride, step, or stance durations during jogging (Lee, Mellifont, & Burkett, 2010) and running (Auvinet, Gloria, Renault, & Barrey, 2002; Wixted, Billing, & James, 2010). To the authors' knowledge, no study has focused on using an IMU to estimate trunk inclination during the sprint start.

The quality of trunk motion estimates provided by inertial sensors has been assessed for clinical applications only during daily-life activities or under controlled flexion-extension, lateral bending and torsion (Lee, 2003; Goodvin, Park, Huang, & Sakaki, 2006; Plamondon, et al., 2007; Wong, & Wong, 2008; Faber, Kingma, Bruijn, & van Dieen, 2009). The results of these studies cannot be directly extended to sprint start, since the quality of the IMU estimates strictly depends on the amount and variation of acceleration entailed in the analysed motor task for the following reasons. First, skin-mounted sensors result in significantly greater peak accelerations than bone mounted sensors (Lafortune, 1991; Forner-Cordero, et al., 2008) and this is particularly true for movements characterised by high accelerations and in areas where soft tissues may undergo increased wobbling (Liu & Nigg, 2000; Pain & Challis, 2006). Second, it is known that the accuracy of sensor orientation estimate can be improved using adaptive algorithms for weighting the acceleration or the angular velocity signal (Sabatini, 2006), with the former prevailing during static or quasi-static phases and the latter during jerked phases, typical of sprint start. Last but not least, such prevalence of angular velocity signals entails dealing with the relevant drift which causes errors that accumulate over time when estimating the angular displacement through numerical integration (Woodman, 2007).

Within this framework, the purpose of this study was to provide coaches with an instrument that can be reliably used in the field and that is able to provide information about the inclination of the trunk, through a three step approach:

1. to minimize problems related to IMU fixing, data fusion, and drift;
2. to assess the accuracy that characterises the estimate of the rotation about a local axis of an IMU;
3. to quantify the relationship between the latter rotation and the instantaneous inclination of the trunk in the plane of progression during both the block start and the pick-up phase.

To these aims, sprint starts were performed in a laboratory. Data provided by an IMU fixed on the lower part of the trunk were compared to reference measurements provided by a stereophotogrammetric system.

3.2.2 Materials and methods

Experimental set up and data acquisition

After institutional approval of the protocol, five male sprinters (age: 23.8±0.8 years; body mass: 72.4±3.8 kg; stature: 1.79±0.07 m) were recruited to take part in the study and gave their written informed consent. The subjects were currently competing over 100 m or 200 m and their best time for 100 m was 11.35±0.15 s. These times are representative of trained college level athletes.

Each sprinter was asked to perform four starts from the starting blocks. All sprint runs were performed in an indoor lab (12x9x4m). The blocks were directly fixed to a wooden tile, which replaced one of the regular floor tiles, and were individually set by the athletes. The two main body positions during the block start phase, “on your marks” and “set” positions, and the first three steps of each start were analysed. To avoid fatigue, a five minutes rest period was given between trials. Each subject was equipped with an IMU (FreeSense, Sensorize Ltd, Italy) containing a 3D accelerometer (±6 g of full range) and a 3D gyroscope (±500 $deg \cdot s^{-1}$ of full range) providing 3D accelerations and angular velocities with respect to a local sensor-embedded system of reference coinciding with the geometrical axes of the IMU case. IMU data, whose sampling frequency was set at 100 frames per second, were sent via Bluetooth® to a laptop computer and low-pass filtered using a moving local regression on windows of 40 samples based on weighted linear least squares and a second degree polynomial model (*smooth* function with the *loess* method, v7.9, MathWorks®, USA). An IMU was positioned with an elastic belt on the lower back trunk (L2 level). In order to limit the wobbling of muscular and soft tissue masses (Forner-Cordero, et al., 2008), a memory foam material placed between the paravertebral muscles and the elastic belt (Fig. 1). The IMU x and y axes (Fig. 2) were aligned with the spine and with the line joining the posterior superior iliac spines, respectively.

To validate the use of the IMU, a nine camera stereophotogrammetric system (Vicon MX3, Oxford, UK) was used. Four retro-reflective markers (LI, LS, RI, RS) were attached to the IMU to determine the unit orientation in space (Fig. 1). Ten markers were positioned on the trunk (right and left shoulder, RSHO and LSHO, spinous process of the 7th Cervical Vertebrae, C7, 10th Thoracic Vertebrae, T10) and on the pelvis (right and left posterior superior iliac spine, RPSIS and LPSIS, and anterior superior iliac spine, RASIS and LASIS) (Fig. 1).

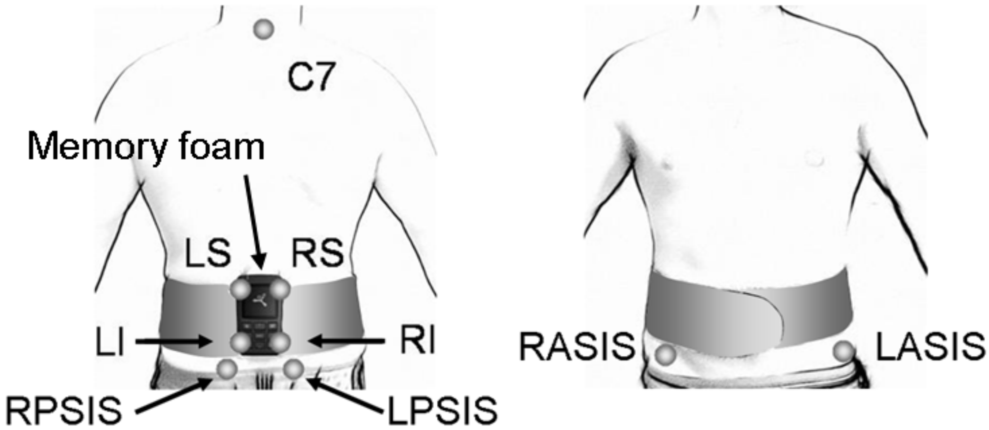


Figure 1 (3.2): Markers and IMU placement. Four retro reflective markers (LI, LS, RI, RS) were attached on the IMU. Ten markers were positioned on the trunk: right and left shoulder (RSHO, LSHO), spinous processes of the 7th Cervical Vertebrae (C7), 10th Thoracic Vertebrae (T10), right and left posterior superior (RPSIS, LPSIS), and anterior superior iliac spines (RASIS, LASIS). Further markers were placed to ease the reconstruction and labeling procedure. Indication of the memory foam material location is provided.

Data processing

IMU angular velocity and acceleration were measured in the IMU Technical reference frame (**T**IMU). An inertial reference frame was defined in a static phase preliminary to the acquisition aligning the x-axis of the **T**IMU with the gravity vector (IMU Global reference frame – **G**IMU). The 3 by 3 rotation matrix, which expressed the orientation of **T**IMU with respect to **G**IMU (${}^{G_{IMU}}\mathbf{R}_{T_{IMU}}$), was then computed (Fig. 2).

To improve the accuracy of the ${}^{G_{IMU}}\mathbf{R}_{T_{IMU}}$ estimate, a Kalman algorithm (Kalman, 1960; Jurman, et al., 2007) was designed to automatically identify the static and non-static phases of the movement and to use a proper combination of the information provided by the accelerometer and gyroscope sensors. Briefly, the filter works as follows:

- when the unit does not move or moves at constant velocity, i.e. the norm of the acceleration vector measured by the unit is below a defined threshold (*s*), its inclination relative to the gravitational acceleration is computed through a quaternion based approach (Favre, Jolles, Siegrist, & Aminian, 2006);

- in all other circumstances the inclination of the IMU relative to gravitational acceleration is estimated by integrating the angular velocity signal, with the initial conditions obtained by accelerometer-based estimates.

Since the analysed movement was supposed to occur in the average plane of progression, only angular velocities measured around the medio-lateral axis were integrated. The importance of correctly selecting the Kalman algorithm parameters has been recently underlined in the literature (Luinge, Veltnik, & Baten, 1999; Donati, Mazzà, McCamley, Picerno, & Cappozzo, 2010). On the basis of pilot trials and of the electronic characteristics of the unit sensors, the following filter parameters were defined for the threshold (*s*) and for static noises (*n*) both for the accelerometer and the gyroscope: $s = 0.5 \cdot g\ m \cdot s^2$ and $n_{GYRO} = 1e^{-008}\ deg \cdot s^{-1}$, $n_{ACC} = 1e^{-009}\ m \cdot s^{-2}$, where g is the norm of the gravitational acceleration.

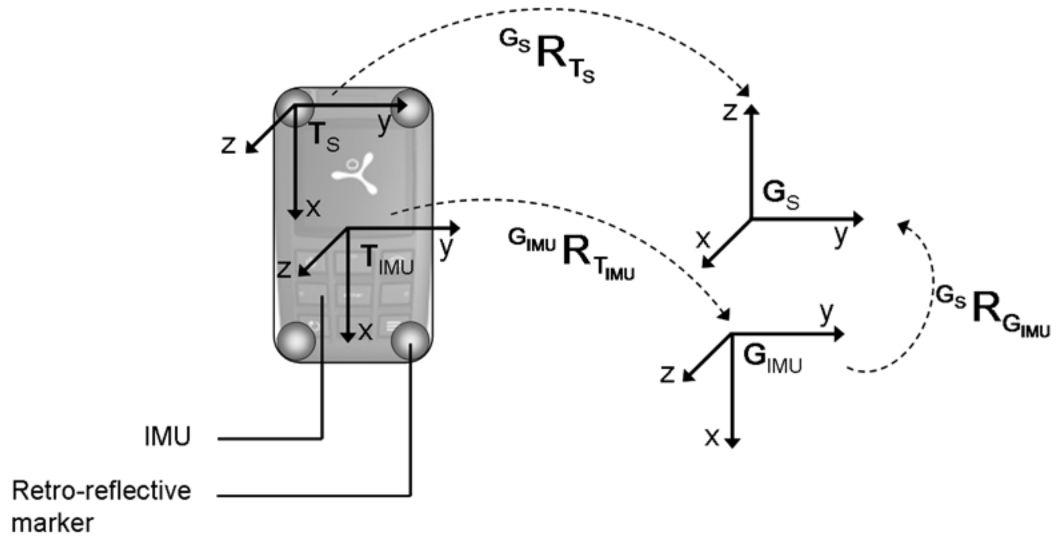


Figure 2 (3.2): Technical (T) and Global (G) frames for the inertial sensor (IMU) and the stereophotogrammetric system (S).

${}^{G_S}\mathbf{R}_{T_S}$: orientation of the four marker frame with respect to the Stereophotogrammetric Global reference frame;

${}^{G_{IMU}}\mathbf{R}_{T_{IMU}}$: orientation of the technical IMU frame with respect to the IMU Global reference frame;

${}^{G_S}\mathbf{R}_{G_{IMU}}$: orientation of the IMU Global reference frame with respect to the Stereophotogrammetric Global reference frame.

In order to assess the accuracy with which the orientation of the IMU technical frame in the global frame was estimated using the above-described procedure, the same orientation was determined using stereophotogrammetric data. A stereophotogrammetric technical reference frame was defined from the four markers placed on the unit (**T**$_S$). This frame was assumed to be coincident with **T**$_{IMU}$ in the first instant of the trial ($^{T_{IMU}}\mathbf{R}_{T_S} = \mathbf{I}$). The orientation of **T**$_S$ with respect to the stereophotogrammetric global reference frame (**G**$_S$) ($^{G_S}\mathbf{R}_{T_S}$) was then obtained in each instant of time. The time invariant rotation matrix relating the two global reference frames, **G**$_{IMU}$ with respect to **G**$_S$, was calculated as follows:

$$^{G_S}\mathbf{R}_{G_{IMU}} = {}^{G_S}\mathbf{R}_{T_S} \cdot {}^{T_S}\mathbf{R}_{G_{IMU}} \tag{1}$$

The orientation of the unit in the stereophotogrammetric global reference frame, **T**$_{IMU}$ with respect to **G**$_S$, in each instant of time was then obtained:

$$^{G_S}\mathbf{R}_{T_{IMU}} = {}^{G_S}\mathbf{R}_{G_{IMU}} \cdot {}^{G_{IMU}}\mathbf{R}_{T_{IMU}} \tag{2}$$

Finally, Tait–Bryant angles were calculated for the orientation of the unit in the reference global frame as measured from stereophotogrammetry and IMU ($^{G_S}\mathbf{R}_{T_S}$ and $^{G_S}\mathbf{R}_{T_{IMU}}$, respectively), using the axis mobile rotation sequence yxz (see Fig. 2 for axes orientation). This rotation sequence was motivated by the fact that the largest rotations occurred about the y-axis. Only the rotation about this axis, here referred to as "pitch angular displacement", was further considered.

In order to synchronize the IMU with the stereophotogrammetric data, a trunk flexion-extension was performed by each subject before each trial. The angular velocity around the y-axis obtained by the stereophotogrammetric data (de Leva, 2008) as well as by the IMU measurements were compared for each trial. The instant corresponding to the peak of angular velocity during the trunk flexion-extension was considered as the first frame of the trial for both systems.

To compare the IMU inclination to the whole trunk inclination, a trunk anatomical frame (**A**$_{TR}$) was defined as follows: *O:* midpoint between LPSIS and RPSIS, *x*-axis joining C7 and *O*, positive downward, *z*-axis directed as the vector product between the *x*-vector and the vector LASIS-RASIS, *y*-axis orthogonal to the *x-z* plane positive towards the left.

The pitch angular displacement relative to $\mathbf{G}_S$ (β) was obtained from the IMU, ${}^{G_S}\mathbf{R}_{T_{IMU}}$, from the unit markers, ${}^{G_S}\mathbf{R}_{T_S}$ and from the whole trunk, ${}^{G_S}\mathbf{R}_{A_{TR}}$. Four phases were identified on the pitch angular displacement (on your marks = OYM, transition = TNS, set = SET, pick-up = PICK-UP phases) separating static and non-static phases (Fig. 3). The following parameters were computed for each curve:

- β_{OYM} and β_{SET}: average values during OYM and SET phases, respectively;
- $\Delta\beta$: difference between β_{SET} and β_{OYM};
- ω_{TNS}: peak angular velocity during the TNS phase;
- $\omega_{PICK\text{-}UP}$: peak angular velocity during the PICK-UP phase.

The offset between the pitch angular displacements calculated using ${}^{G_S}\mathbf{R}_{T_{IMU}}$ and ${}^{G_S}\mathbf{R}_{A_{TR}}$ was calculated in the OYM position (β_{OFF}).

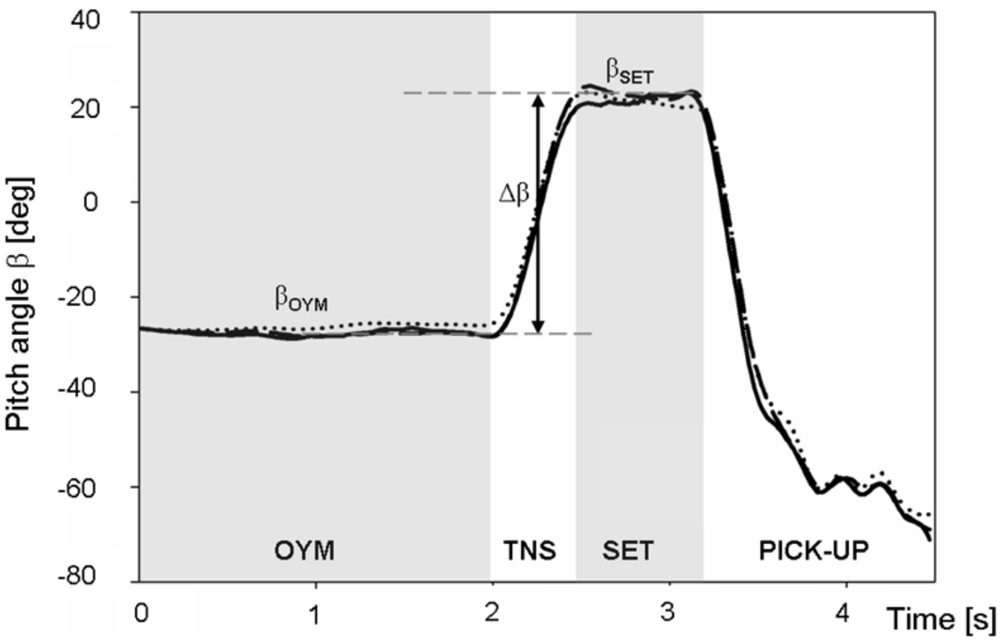


Figure 3 (3.2): Typical pitch angles (β) for one trial as obtained from the IMU technical frame (solid line), from the stereophotogrammetric technical frame (dotted line) and from the trunk anatomical reference frame (dashed line). The angular displacement parameters estimated for each phase (OYM: on your marks, TNS: transition, SET: set, PICK-UP: pick-up) are indicated: β_{OYM} and β_{SET}: average values of the pitch angular displacement during the OYM and SET positions, respectively; $\Delta\beta$: peak to peak difference between β_{SET} and β_{OYM}. The pitch angular displacement was considered to be zero when the unit was in a horizontal position; positive angles correspond to clockwise rotations.

Statistical analysis

To assess the accuracy of the parameters provided by the IMU against the $\mathbf{T}_S$ reference measurements, as well as the relationship between the information provided by the IMU and the trunk considered as a rigid segment ($\mathbf{A}_{TR}$), a two steps statistical analysis was performed.

First, curve similarity was evaluated between pitch angular displacements obtained from $\mathbf{T}_{IMU}$ and $\mathbf{T}_S$, for IMU accuracy assessment, and from $\mathbf{T}_{IMU}$ and $\mathbf{A}_{TR}$, for IMU-trunk rotation consistency, computing the root mean square difference (RMSD) and the Pearson's product-moment correlation coefficient (r). It has to be underlined that, when dealing with IMU accuracy, the RMSD between the unit estimates and reference measurements can be interpreted as root mean square error. The same coefficients were also assessed for only the PICK-UP phase, potentially the most affected by inertial factors.

Second, for each computed parameter, the following statistical analysis was performed:

1. Descriptive statistics (mean±standard deviation);
2. Mean bias analysis: difference between the stereophotogrammetric system parameters (S) and IMU measurements (IMU);
3. Investigation of the presence of a linear trend between the amount of random error and the measured values (heteroscedasticity) through inspection of Bland and Altman plots and correlation analysis (Nevill & Atkinson, 1997).

3.2.3 Results

The comparison of pitch curves, obtained from $\mathbf{T}_{IMU}$ and $\mathbf{T}_S$ when assessing the accuracy of the IMU, resulted in a low root mean square difference (RMSD) and a very high correlation (r) during block and pick-up phases (Tab. 1). Similar results were obtained for IMU-trunk rotation consistency, comparing pitch curves obtained from $\mathbf{T}_{IMU}$ and $\mathbf{A}_{TR}$ (Tab. 1).

	IMU	TR
r	0.994 ± 0.013	0.998 ± 0.002
$r_{PICK-UP}$	0.995 ± 0.015	0.998 ± 0.001
RMSD [deg]	3 ± 2	3 ± 2
$RMSD_{PICK-UP}$ [deg]	3 ± 3	3 ± 2

Table 1 (3.2): Curve similarity: Mean ± one standard deviation of the Root Mean Square Difference (RMSD) and the correlation coefficient (r) computed to assess unit accuracy (IMU) and IMU-trunk rotation consistency (TR), relative to the whole task and to the PICK-UP phase.

For all pitch parameters, bias was lower than 1 deg, for the IMU accuracy ($bias_{IMU}$), and than 4 deg, for IMU-trunk rotation consistency ($bias_{TR}$). The offset of the pitch angle between the unit and the trunk reference frame, computed in the OYM phase, showed a low variability among subjects and trials (β_{OFF} = 18±4 deg). The peak bias of the angular velocity, during both the TNS and the PICK-UP phases, was found to be lower than 9 and 6 deg·s^{-1} for IMU accuracy and IMU-trunk rotation consistency, respectively (Tab. 2).

	T_S	T_{IMU}	A_{TR}	T_S VS T_{IMU}	A_{TR} VS T_{IMU}
				$bias_{IMU}$	$bias_{TR}$
β_{OYM} [deg]	-29±6	-29±6	-29±6	1 ± 1	0 ± 1
β_{SET} [deg]	12±6	12±7	9±6	1 ± 1	4 ± 4
$\Delta\beta$ [deg]	43±7	43±8	39±5	1 ± 1	4 ± 4
ω_{TNS} [deg·s^{-1}]	106±19	98±18	93±18	8 ± 4	4 ± 6
$\omega_{PICK-UP}$ [deg·s^{-1}]	-201±25	-192±29	-198±31	9 ± 6	6 ± 13

Table 2 (3.2): Parameter analysis: Descriptive statistics (mean ± standard deviation) of each estimated parameter and of the errors between the estimates of technical stereophotogrammetric frame (T_S), the IMU frame (T_{IMU}), and the anatomical trunk frame (A_{TR}). β_{OYM} and β_{SET}: average values of the pitch angular displacement during the OYM and SET positions, respectively; $\Delta\beta$: peak to peak difference between β_{SET} and β_{OYM}; ω_{TNS}: peak angular velocity during the TNS phase; $\omega_{PICK-UP}$: peak angular velocity during the PICK-UP phase.

Symmetry of the confidence interval of the bias was observed for all parameters listed in Tab. 2, during visual inspection of the Bland and Altman plots entailing that no systematic error was present. No linear trend between the amount of bias and the measured values was found (average correlation = 0.09).

3.2.4 Discussion

The accuracy of an IMU positioned on the lower part of the trunk in estimating its rotation about a local axis during sprint start, as well as the relationship between the IMU estimates and the athletes' trunk inclination were assessed. The results show that problems associated with the IMU can be overcome so that coaches can be provided with an instrument that can be reliably used in the field to characterise the inclination of the trunk considered as a rigid body.

The accuracy of the IMU is shown by the high agreement between the IMU and the reference measures, during the whole trial as well as during the PICK-UP phase, for both curve similarity and parameter analysis. Such agreement proves that the main errors entailed in using IMUs to estimate inclination during dynamic exercises did not have a disruptive effect on the final results. First, the proper identification of static and non-static phases, as well as the selection of the weighting on the accelerometer and the gyroscope measurements, are critical to estimate the unit orientation using Kalman filtering (Sabatini, 2006). The efficacy of the used filter is attested by the high curve similarity obtained during a combination of static and non-static phases, as well as during the pick-up phase. Curve similarity allows hypothesizing that the unit fixing site and method (elastic belt plus memory foam material placed at L2 level) are a good solution in order to limit the artefact introduced by soft tissue oscillations. Last but not least, the drift errors entailed in the numerical integration process of the angular velocity signals (Woodman, 2007) proved to be negligible, given the short duration of the integration interval (from the SET end to the upright position there was less than 1 s).

The trunk inclination, considered as a rigid segment, and the IMU inclination were in high agreement during the whole curve and the PICK-UP phase. The trunk average peak angular velocity ($\mathbf{A}_{TR}$: $\omega_{PICK\text{-}UP}$ = 198 ± 31 deg·s^{-1}) agrees with previous literature results, although slightly higher (186 ± 48 deg·s^{-1} in Natta et al., (2006), and about 150 deg·s^{-1} in Slawinski et al., (2010). This difference can be explained considering that both studies analysed the task on a real track, where it is plausible that athletes tended to reach an upright position more gradually with respect to the athletes tested in the present study, who were forced to complete the task before reaching the end of the lab. The sensor and the trunk pitch angles in the OYM phase presented an offset due to a different alignment of the IMU and the trunk, depending on subjects' anatomy and due to the athletes' kyphosis. The low values of RMSE and the relevant high correlation indicate, consistently between subjects, that the initial offset does not change during all phases. It could be speculated that a calibration between the unit and the athlete trunk should not be necessary, whenever the attention is focused on the variation of the angular displacement, as it is often the case in coaches' perspective.

The agreement between the unit and the trunk inclination seems to support track and field coaches' approach in considering the trunk as a rigid segment. From the expert coaches' perspective, the orientation of the trunk is usually observed only in the progression plane, as the angular displacement is preponderant in this plane, and it is considered as a rigid segment (Jones, et al., 2009). Although it is well known that the trunk has a complex behaviour during the sprint, in the light of the previous considerations, it is questionable whether a more complex analysis would be of any use to a coach used to qualitatively observing the task.

In conclusion, the study shows that a single IMU positioned on the lower back trunk provides reliable angular displacements in the plane of progression during a sprint start from blocks. Moreover, from these estimates, it is possible to quantify the inclination of the trunk considered as a rigid segment. The inclination of the trunk could be particularly helpful for trainers, being one of the key elements in block start and pick-up performances. Future works, will therefore, concern the validation of the method in-field, in order to analyse professional sprinters and all-out sprint starts, exploring, in particular, the correlation between the estimated parameters and the whole race performance.

3.3 STUDY 2: INSTANTANEOUS VELOCITY AND CENTER OF MASS DISPLACEMENT OF IN-LAB SPRINT RUNNING USING AN INERTIAL MEASUREMENT UNIT

Abstract

The biomechanical analysis of sprint running during in-field training provides valuable information regarding athlete motor strategies, helping in preventing injuries and achieving higher athletic performance. Wearable inertial sensing units (IMUs) providing movement-related data without any space limitation or cumbersome set up can be proficiently used to perform such analysis. Nevertheless the noise characterizing these sensor signals exacerbates the process of determining velocity and position by numerical integration of acceleration.

This study aims at compensating these errors by cyclically determining the initial conditions of the integration process to yield reliable instantaneous velocity as well as spatio-temporal parameters during sprint running.

Six sprinters performed three in-lab sprint runs, starting from a standing position. The stance time (ST), and the progression displacement (d) and mean progression velocity (v) of a point approximating the centre of mass were estimated and compared with reference data (force platform and stereophotogrammetric measurements).

Preliminary results showed a high correlation ($r > 0.9$) between IMU and reference estimates for each parameter. No statistical differences were found between IMU and reference for v and ST.

These results proved that the methodology successfully compensate the numerical integration errors during non steady-state running. Future developments will concern in-field sprint running experimental sessions.

Keywords: Sprint running; Velocity; Displacement; Inertial sensors; Biomechanics.

3.3.1 Introduction

Field performance analysis of sprint running can be carried out using several approaches, according to the inquired resolution. Total running time and step frequency are basic yet very useful information for a global assessment of the athlete's performance (Mehrikadze & Tabatschnik, 1982).

Inertial sensors have been used to determine these or similar simple parameters. Foot-ground contact times (Auvinet, et al., 2002; Purcell, et al., 2005; Wixted, et al., 2010) and lower leg rotational kinematics (Channells, et al., 2005) have been assessed using accelerometers on the tibia. Displacement measurements and running speed have been estimated placing IMUs on the foot (Fyfe & Gildenhuys, 2004) and, during steady-state running, in a chest-belt (Vetter, Onillon, & Bertschi, 2009), respectively.

However, when the focus is on performance determinants and limiting factors, a deeper analysis could be necessary, so that to provide track and field coaches with detailed information about the athlete's technique. Stance duration, step length and center of mass (CoM) instantaneous velocity are key quantities in such analysis. Theoretically, the numerical integration of the sensor-measured acceleration yields to the instantaneous velocity and displacement of the segment where the IMU is attached. In practice, the numerical integration process remains challenging due to two sources of error. First, accelerometers are characterised by offset errors that rapidly accumulate over time and yields to unreliable velocity and displacement (Woodman, 2007). Although in constant speed running, the drift of the velocity is linear and can be removed with a filtering procedure (Pfau, Witte, & Wilson, 2005; Pereira, et al., 2008), during non steady-state running, like sprinting is, this drift is no longer linear. Second, when dealing with explosive movements, sensor wide oscillations caused by the inertia of soft tissues (de Leva & Cappozzo, 2006) and by the fixing device (Forner-Cordero, et al., 2008) lead to sudden velocity increase/decrease that could highly under/overestimate the actual velocity.

In this pilot work, a methodology for estimating instantaneous velocity and displacement of a point approximating the CoM from trunk accelerometry during non steady-state running is illustrated. Low frequency errors are compensated by reducing the numerical integration interval to the duration of the stance phase and by predicting the kinematics of the sensor during the flight phase. The initial conditions of the integration process are, then, cyclically determined.

3.3.2 Materials and methods

Experimental set up and data acquisition

Six sprinters (five males: age = 27±2 yrs, stature = 1.78±0.07 m, mass = 72±7 kg; one female: age = 27 yrs, stature = 1.65 m, mass = 65 kg) gave their written consent to participate in this study. After 20 minutes warm up, each athlete was asked to perform three sprint runs, starting from a standing position. To avoid fatigue, a 10 minutes rest period was given between trials.

Due to limited laboratory volume (12x9x4m) only the first three steps were analysed. Subjects were equipped with 39 markers to determine the instantaneous 3D position of the CoM (Plug-in Gait protocol - Davis et al., 1991) using a nine-cameras stereophotogrammetric system (Vicon Mx, Oxford, UK). Two six-component force plates (Bertec) were used to measure ground reaction forces at the second and the third steps.

According to preliminary tests, an IMU (MTx, Xsens, Netherlands) was positioned using a specifically designed shoulder-belt on the upper back trunk (C7 level). This sensor embedded 3D accelerometer, 3D gyroscope and a magnetometer. Through the combination of the information provided by these sensors, the orientation of the IMU relative to the magnetic North and to the gravity line could be obtained. Stereophotogrammetry, force platforms, and inertial sensors data were collected simultaneously at 100 samples per second.

Data processing

Before estimating velocity and displacement, accelerations were expressed with respect to an inertial reference frame (M), aligned with the sprinting direction (${}^{M}\mathbf{a}$) (Cappozzo, della Croce, Leardini, & Chiari, 2005). Assuming the trunk as a rigid segment, the acceleration of the IMU was translated along the longitudinal axis to the mid point between posterior superior iliac spines (midPSIS), closer to the CoM. Thereafter, the instantaneous progression velocity and displacement were computed by numerical integration of the acceleration. The integration was limited to the stance phase only, to avoid the drift typical of the integration process. During the flight phase, the horizontal kinematics of midPSIS was predicted using the ballistic law of motion; the velocity at the instant of take-off was set to the last value of the previously integrated acceleration. This procedure was reiterated for each step.

Foot contact times were determined using the vertical component of ${}^{M}\mathbf{a}$. Once having identified the highest high-frequency peaks in the time-derivative of the acceleration signal, the beginning and the end of the flight phase were determined moving downward through the signal until the acceleration approached ≅ $-9\ m \cdot s^{-2}$.

The following parameters were estimated over each step: 1) stance time ($\mathbf{ST_e}$); 2) CoM progression displacement ($\mathbf{d}_e$); 3) mean progression velocity ($\mathbf{v}_e$) computed as the arithmetic mean of the instantaneous progression velocity between two consecutive foot strikes. Reference data were obtained as follows: $\mathbf{ST_r}$ from force plates (steps 2-3); $\mathbf{d}_r$ and $\mathbf{v}_r$ from the CoM trajectory obtained by stereophotogrammetry. The average percentage difference ($e_{\%}$) between IMU and reference estimates, referred to as error, was calculated for each parameter.

Statistical analysis

Statistical analysis was performed using SPSS (version 17.0), with α set to 0.05. Multivariate ANOVA was performed across the three performed trials on the estimated parameters to establish whether significant differences existed among trials. Two-tailed paired t-test was performed to establish whether significant differences existed between methods (reference and estimate). Pearson's correlation coefficient was computed on reference and estimated **d** and **v** values. Least significant difference post hoc comparisons with Bonferroni adjustments were used for multiple comparisons.

3.3.3 Results

The methodology developed in this study yielded to more reliable instantaneous progression velocity with respect to that obtained by standard numerical integration of the acceleration (Fig. 1). No differences were found for **ST**, **d** and **v** values across the three trials ($p > 0.05$). No differences ($p > 0.05$) and high correlation ($r = 0.957$) were found between the sensor and the reference estimates of the velocity ($\mathbf{v}_e$ and $\mathbf{v}_r$). Similarly, no differences ($p > 0.05$) were found between $\mathbf{ST_e}$ and $\mathbf{ST_r}$ over the first and the second stance phases. Despite statistical differences were found between $\mathbf{d}_e$ and $\mathbf{d}_r$ ($e_{\%}$ = 10%), a high correlation was found between these parameters ($r = 0.967$). Correlation between sensor and reference estimates for **d** and **v** is shown in Fig. 2.

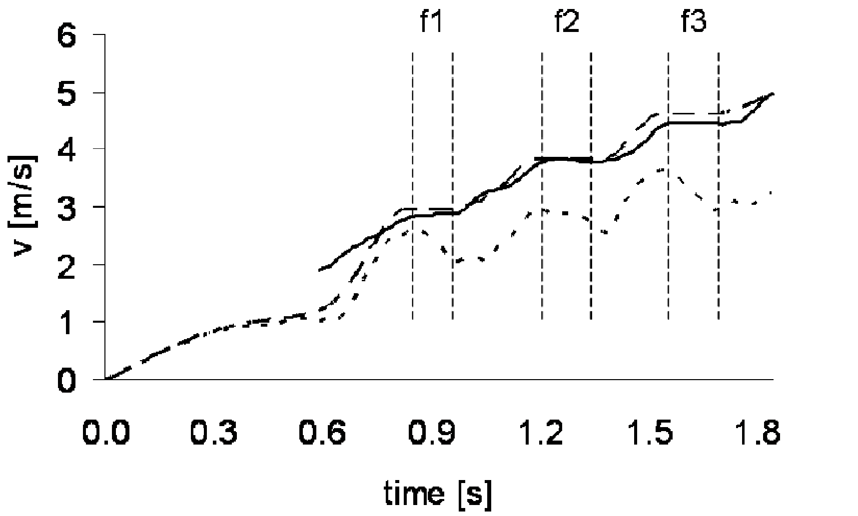


Figure 1 (3.3): Instantaneous progression velocity as obtained by the reference measurements (solid line), computed by numerical integration of the acceleration for the whole duration of the task (dashed line) and with the algorithm proposed in this study (dashed line). Vertical dashed lines identify the flight phases (f1, f2 and f3).

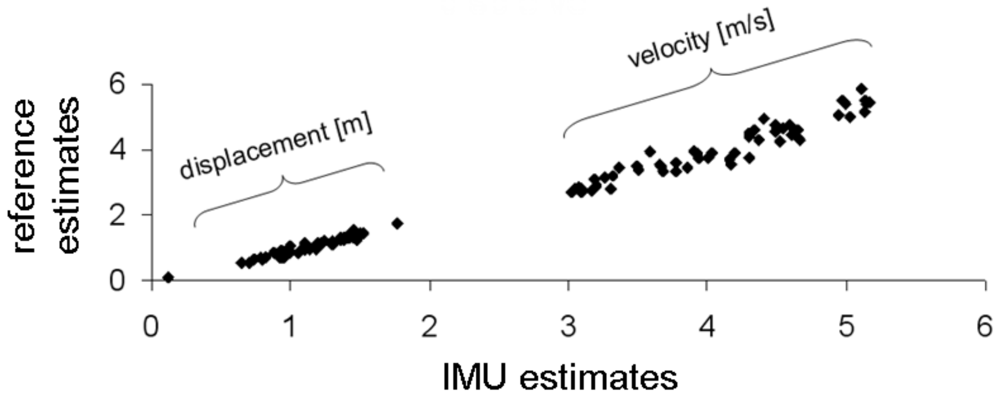


Figure 2 (3.3): Reference and estimated progression displacement and mean velocity of the CM for all the collected trials.

3.3.4 Discussion

The investigation of performance determinants and limiting factors of sprint running requires the estimate of the linear kinematics of the CoM of the subject. When the analysis is performed using inertial sensors, the kinematics of the CoM can be estimated only through assumptions and simplified models. In the present study, errors introduced by the numerical integration of the acceleration were compensated by limiting the integration to the stance phase and by predicting the kinematics of the sensor during the flight phase using the ballistic law of motion. The initial conditions of the integration process were, then, cyclically determined. The instantaneous horizontal velocity and displacement of the CoM were estimated.

Results concerning the reliability of the estimates showed a high agreement between the sensor and the reference measurements for each estimated parameters, and a high accuracy for **v** and **ST**. Significant differences were found between reference and IMU estimates only for **d**. This is probably due to the double integration process necessary to obtain **d** from the acceleration. Despite such differences, the mean error was found to be lower than 10% with respect to the reference measurements. Moreover, as above mentioned, a high correlation between reference and IMU **d** estimates was found (Fig. 2). It has to be noted that **d** is strictly related to the athlete's step length, a very important parameter in performance assessment which is of great interest for track and field coaches. However, further research dealing with this correlation is necessary.

Results concerning the estimates of the instantaneous velocity (**v**) proved that the drift error entailed in the numerical integration of the acceleration signal was minimised. The shoulder belt used for the fixation of the IMU, aiming at reducing the movement of the soft tissues relative to the skeleton with respect to the pelvic area, appears to be a reasonable compromise between methodological and practical requirements. Nevertheless, athletes' arms oscillation during the acceleration phase was partially disturbed by the underarm stripes. Therefore, further investigations about the inertial unit fixation have to be performed.

In conclusion, the proposed methodology proved to compensate numerical integration errors during non steady-state running. The agreement found between the measured and estimated parameters showed that the methodology is sensitive to the variations of these parameters. Aside from the encouraging results obtained in this study, in-field validation of the proposed method is necessary.

3.4 <u>STUDY 3</u>: TEMPORAL PARAMETERS OF IN-FIELD SPRINT RUNNING USING AN INERTIAL MEASUREMENT UNIT

Abstract

Sprint running temporal parameters are useful information for track and field coaches. Using wearable inertial measurement units (IMUs) to perform an in-field analysis of these variables requires being able to temporally locate events correlated with foot-ground contacts by identifying consistent and repeatable features on the signal waveforms measured by the IMU. Feature-identification, however, highly depends on the signal to noise ratio, especially unfavourable during sprint running analysis because of the explosiveness of the task and of the oscillations of the IMU relative to the underlying skeleton.

The aim of this study was to identify recognizable and consistent features in the waveform of the signals supplied by a trunk mounted IMU, or thereof derived, for the estimation of the stance duration during sprint running.

Eleven subjects (six amateur and five elite athletes) performed three 60 m sprint runs and data provided by an IMU positioned on the trunk at L2 level were compared to reference forceplate and high-speed camera measurements. Feature identification was performed on the magnitude of the acceleration (a) and angular velocity (ω) vectors and on their 1st ($\dot{a}$ and $\dot{\omega}$) and 2nd ($\ddot{a}$ and $\ddot{\omega}$) wavelet-mediated derivatives.

Repeatable features were not identified in either a or its derivatives, while ω was characterized by a consistent positive peak that allows for the identification of the stride cycle duration. The beginning and end of the stance were identified from positive and negative peaks on $\ddot{\omega}$, consistently across subjects and trials. Mean errors in the estimate of the stance and stride phase duration were found to be of the same order than the temporal resolution of the IMU (0.005 s).

Keywords: Sprint running; Contact times; Inertial sensor; Feature-identification; Biomechanics.

3.4.1 Introduction

Temporal parameters, such as flight and stance duration, are basic yet very useful information for track and field coaches. The influence on sprint running performance of these durations and of related parameters, such as stiffness and ground reaction force impulse, has been widely investigated in the literature (Butler, Crowell, & Davis, 2003; Hunter, Marshall, & McNair, 2004a; Hunter, Marshall, & McNair, 2004b; Hunter, Marshall, & McNair, 2005; Morin, Dalleau, Kyrolainen, Jeannin, & Belli, 2005; Čoh, Tomažin, & Štuhec, 2006; Ito, Ishikawa, Isolehto, & Komi, 2006; Ciacci, Di Michele, & Merni, 2010).

Commonly, to compute or estimate temporal parameters during sprint running, force platforms (Hunter, et al., 2004b, 2005), stereophotogrammetric systems (Ciacci, et al., 2010), optical bars (Čoh, et al., 2006) or video analysis (Ito, et al., 2006) have been used. The first two instruments, however, allow for the analysis of few steps, being all limited in terms of acquisition volume. Moreover, as a real sprint run can not be entirely reproduced in a laboratory, it is necessary to bring these instruments on the track, which results to be laborious and time-consuming. Video analysis, which may allow for the analysis of a wider portion of the track, still requires a non automatic post-processing that might be time-consuming as well.

As an alternative to the above mentioned instruments, inertial measurement units (IMUs) have been widely used to measure temporal parameters, particularly during walking (Jasiewicz et al., 2006; Kavanagh & Menz, 2008; Hanlon & Anderson, 2009). Such parameters have been generally determined by identifying mechanically-related features in the acceleration and angular velocity signal waveforms. It is well known, for example, that during walking, the peak forward acceleration coincides with the instant of foot contact (Verkerke, Hof, Zijlstra, Ament, & Rakhorst, 2005) or that a repeatable trunk rotation about the trunk longitudinal axis occurs, during walking as well as during running, prior to foot contact (Schache, Blanch, Rath, Wrigley, & Bennell, 2002; Saunders, Schache, Rath, & Hodges, 2005).

Due to their portability and weightlessness, IMUs have recently gained momentum as a suitable solution for low cost in-the-field biomechanical analysis of running performance as well. So far, only two studies focused on the estimate of foot-ground contact times during sprint running using accelerometer-based systems. Hobara et al. (2009) estimated stance and flight times using a heel-mounted bi-axial accelerometer, for a model-based estimate of the vertical stiffness over a 400 m sprint run. The stance durations were measured from the output waveform of the accelerometer, but no validation was presented nor previously published by the authors. Purcell et al. (2006) validated with a force platform the estimates of stance durations based on a shank-mounted tri-axial accelerometer during the first three steps of the acceleration phase, as well as during the steady-state phase of a sprint running. However, mounting sensors on lower legs, aside from requiring two devices, does not provide information on the body center of mass (CoM) and on trunk kinematics.

These drawbacks may be overcome using a single device on the trunk, which can be reasonably acceptable by runners. Trunk-mounted devices have been used to estimate temporal parameters only during distance running (Auvinet, Gloria, Renault, & Barrey, 2002; Wixted, Billing, & James, 2010). Both authors concluded, on a qualitative level, that the acceleration signals agreed with the waveforms of force platforms (Auvinet, et al., 2002) or of in-shoe pressure sensors (Wixted, et al., 2010), but neither clearly identified mechanically-related features in the signal waveforms nor provided quantitative information on the accuracy of their estimates.

Robustness and reliability of these temporal estimates rely on the accurate and consistent identification of the above-mentioned features within and across subjects which, in turn, highly depends on the signal to noise ratio. In this respect, sprint running analysis is more challenging than walking and distance running because the explosiveness of the task causes greater movements of the IMU relative to the underlying skeleton (Liu & Nigg, 2000; Pain & Challis, 2006; Wakeling & Nigg, 2001). As this artefact was shown to be both subject-dependent (Lafortune, Henning, & Valiant, 1995) and sensitive to the site and method of the unit attachment (Forner-Cordero et al., 2008; Lafortune, 1991), it may jeopardize the consistency of the feature identification approach (Higginson, 2009).

The purpose of this study was, therefore, to identify recognizable and consistent features in the waveform of the signals supplied by a trunk-mounted IMU, or thereof derived, for the estimation of the stride and stance durations during sprint running. To this aim, maximal sprint runs were performed on regular tartan tracks by elite and amateur athletes. Data provided by an IMU fixed on the lower part of the trunk were validated against reference measurements provided by force platforms and a high-speed video camera.

3.4.2 Materials and methods

Experimental set up and data acquisition

Two groups of subjects took part in the study and gave their written informed consent after institutional approval of the protocol. The first group (A) was composed of six amateur athletes: two females and four males. None of them was specialized in track and field disciplines. The second group (B) included five professional sprinters: two women and three men. All of them were currently training and competing in the Italian National Track and Field Team. Anthropometric data of each athlete involved in the study is reported in Tab. 1.

	GROUP A						*GROUP B*				
	A1	**A2**	**A3**	**A4**	**A5**	**A6**	**B1**	**B2**	**B3**	**B4**	**B5**
mass [kg]	56	45	72	60	73	75	56	54	70	73	72
stature [m]	1.71	1.48	1.72	1.83	1.78	1.80	1.70	1.74	1.74	1.80	1.86
personal best [s] (100m)	#	#	#	#	#	#	11.51	11.52	10.17	10.63	10.49

Table 1 (3.3): Mass [kg], stature [m] and personal best (100 m) [s] of the eleven subjects involved in the study.

Two different acquisition sessions were performed: group A was tested on an in-door track at the Institut National du Sport de l'Expertise et de la Performance (Paris, France); while the Italian National Track and Field Team (group B) was tested during one of the scheduled training session for the European Track and Field Championship of Barcelona at the Centro Sportivo Aeronautica Militare (Vigna di Valle, Bracciano, Rome, Italy). The same experimental protocol was used for both groups: after 20 minutes warm up, each athlete was asked to perform three sprint runs of 60 m, starting from a standing position. Four steps at the athlete maximal speed were then analysed. To avoid fatigue, a 10 min rest period was given between trials.

Each subject was equipped with an IMU (FreeSense, Sensorize Ltd, Italy) containing a 3D accelerometer (±6 g of full range) and a 3D gyroscope (±500 deg·s^{-1} of full range) providing 3D accelerations and angular velocities with respect to a local sensor-embedded system of reference coinciding with the geometrical axes of the IMU case. IMU data, whose sampling frequency was set at 200 frames per second, were sent via Bluetooth® to a laptop computer (MathWorks® v7.9, USA). Careful attention was paid to the fixation of the IMU to the athletes' body (Forner-Cordero, et al., 2008) to limit the unit oscillations relative to the underlying bone. The IMU was positioned on the lower back trunk (L2 level) with an *ad-hoc* elastic belt, not limiting the athlete's movements. This location was chosen in order to avoid the low lumbar area, more affected by the wobbling of muscular and soft tissue masses. To further limit the unit oscillations due to the paravertebral muscular mass, a memory foam material was placed between the paravertebral muscles and the unit (see Fig. 1 (3.2), § 3.2.2).

To validate IMU results, two different instruments were used: group A data were validated using six force platforms (AMTI, USA) directly embedded in the track and covered with a layer of tartan, not to influence or disturb the athletes while running. The total platform surface was about 6.6 m length per 0.6 m wide. Sampling frequency was set at 200 frames per second. To collect at least four steps on the platform area at the athlete's maximal speed, the starting line was set at 40 m from the beginning of the first platform. Group B data were validated using a high-speed video camera (Casio Exilim EX-F1, Japan) whose sampling frequency was set at 300 frames per second. The camera was positioned at 40 m from the starting line and registered at least four steps for each athlete. For group A, force platforms and IMU data were synchronised with a small hammer by hitting the platform upon which the IMU was positioned. As, above mentioned, data for elite athletes were collected during an official training session, for practical time reasons no accurate synchronisation between the camera and the IMU signals was possible.

Data processing

IMU angular velocity and acceleration were measured in the local IMU reference frame. As the actual orientation of the unit was not known, the norm of both the acceleration (a) and the angular velocity (ω) vectors was computed and further considered. This was done in order to avoid any source of variability which may arise from different sensor positioning or running technique (particularly as concerns trunk orientation) (Patterson & Caulfield, 2010).

The first ($\dot{a}$ and $\dot{\omega}$) and second ($\ddot{a}$ and $\ddot{\omega}$) derivatives of a and ω were calculated using a wavelet-based approach (Jianwen, et al., 2006).

a, $\dot{a}$ and $\ddot{a}$, as well as ω, $\dot{\omega}$ and $\ddot{\omega}$ signal waveforms of group A were synchronized with the vertical ground reaction force (GRF), and repeatable quantifiable events that could be related with the mechanics of the task, such as maxima, minima, or slopes were identified in correspondence with foot-strike (FS) and foot-off (FO) instants. Features adequate for automatic detection of FS and FO were then identified on both group A and B data and, thereafter, used to estimate stance ($\tilde{d}_{stance}$) and stride ($\tilde{d}_{stride}$) durations.

Validation of results thus obtained was performed by comparing the estimates with reference measurements. Reference d_{stance} and d_{stride} values were obtained for group A setting a threshold of 10 N on rising edges and of 25 N on descending edges of the vertical component of the GRF. These thresholds were set as criteria measures for identifying FS and FO instants respectively (Hunter, et al., 2005). As concerns group B, a frame by frame video-analysis was performed and FS and FO instants were visually identified to measure d_{stance} and d_{stride}.

Statistical analysis

To assess the accuracy of $\tilde{d}_{stance}$ and $\tilde{d}_{stride}$ the following statistical analysis was performed separately on group A and group B results:

1. A repeated-measures analysis of variance (ANOVA) was used to verify the effect of the factors groups, subject and trial on the estimated parameters, both for the reference and the IMU values.
2. Descriptive statistics (mean ± standard deviation (SD)) of both $\tilde{d}_{stance}$ and $\tilde{d}_{stride}$;
3. Bland and Altman plots, corrected for the effect of repeated measurement error (Bland & Altman, 2007), were used to assess the agreement between methods (reference and IMU estimates) with multiple observations (Atkinson & Nevill,

1998). The absence of a linear trend between the amount of random bias and the measured values (heteroscedasticity) was investigated through inspection of Bland and Altman plots and correlation exploration (Nevill & Atkinson, 1997).

4. Finally, for each parameter a repeated-measures analysis of variance (ANOVA) was used to verify the effect of the factors group, subject and trial on the bias between reference and IMU estimates.

Statistical analysis was performed using SPSS (version 17.0). The alpha level of significance was set to 0.05.

3.4.3 Results

The mean ± one standard deviation of a (upper panel) and ω (lower panel) signals for group A are presented in Fig. 1. Stride cycles were segmented through the force platform signals and expressed in percentage of the total cycle duration.

No repeatable and quantifiable features, adequate for automatic detection, were identified on a, neither in group A nor in group B. Conversely, the magnitude of the angular velocity signal was characterized by a consistent positive peak (ω_{peak}) which occurred approximately at the end of each step cycle in both groups. This peak was clearly visible even by simple visual inspection of the signal and was used to segment the stride cycle of Group B data.

As concerns the stance phase duration, no repeatable features related to the FS and FO instants were recognizable in either the derivatives of a or in ω (Fig. 2). FS and FO events, indeed, did not occur at consistent maxima, minima, or even slope across different subjects (Fig. 2).

Fig. 3 shows ω inter-subject variability for three selected subjects of both group A and group B.

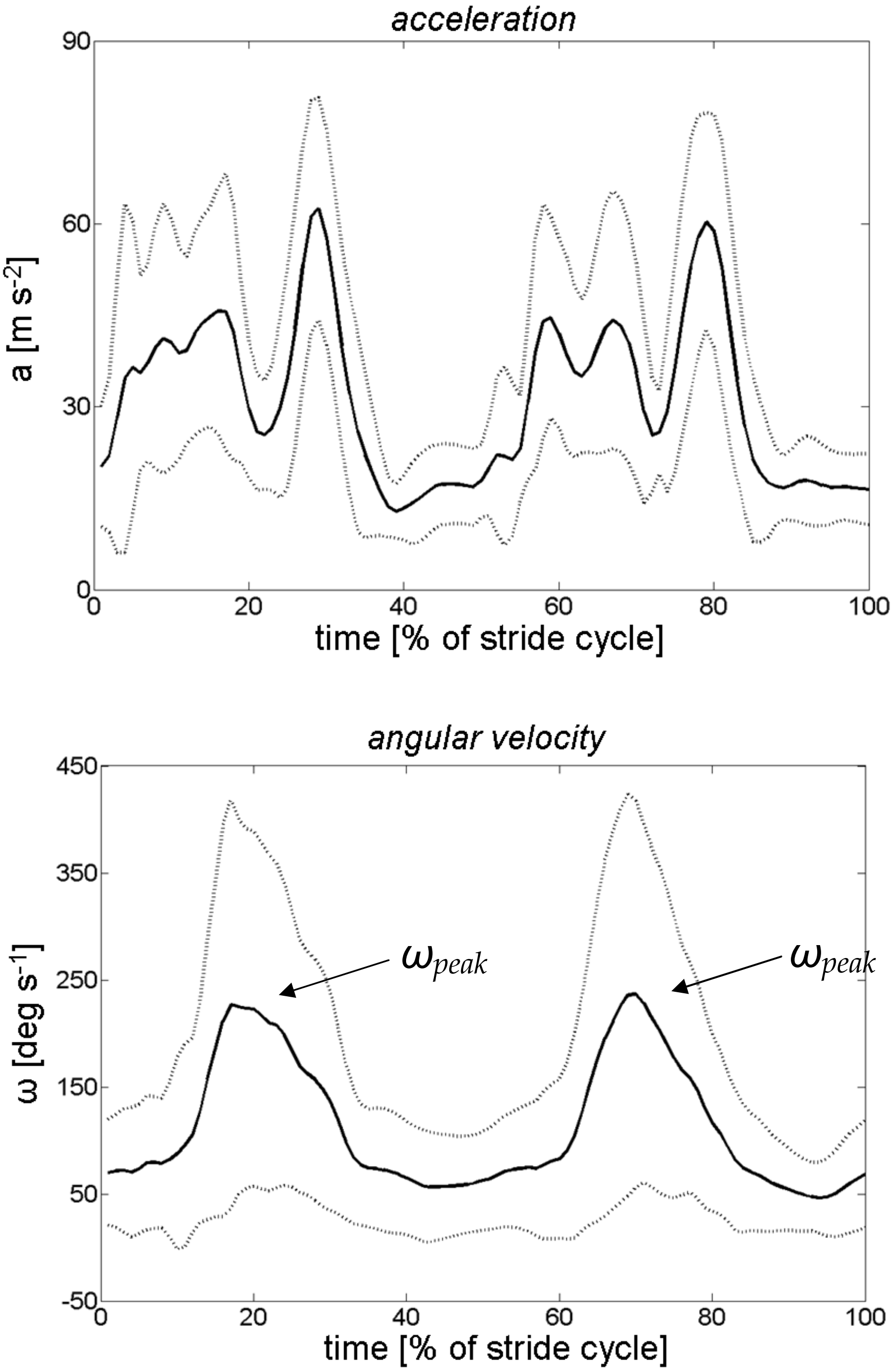


Figure 1 (3.4): Mean ± 1SD of 6 subjects x 3 trials x 4 strides of Group A for a (upper panel) and ω (lower panel). The positive peak of the angular velocity, ω_{peak}, is indicated.

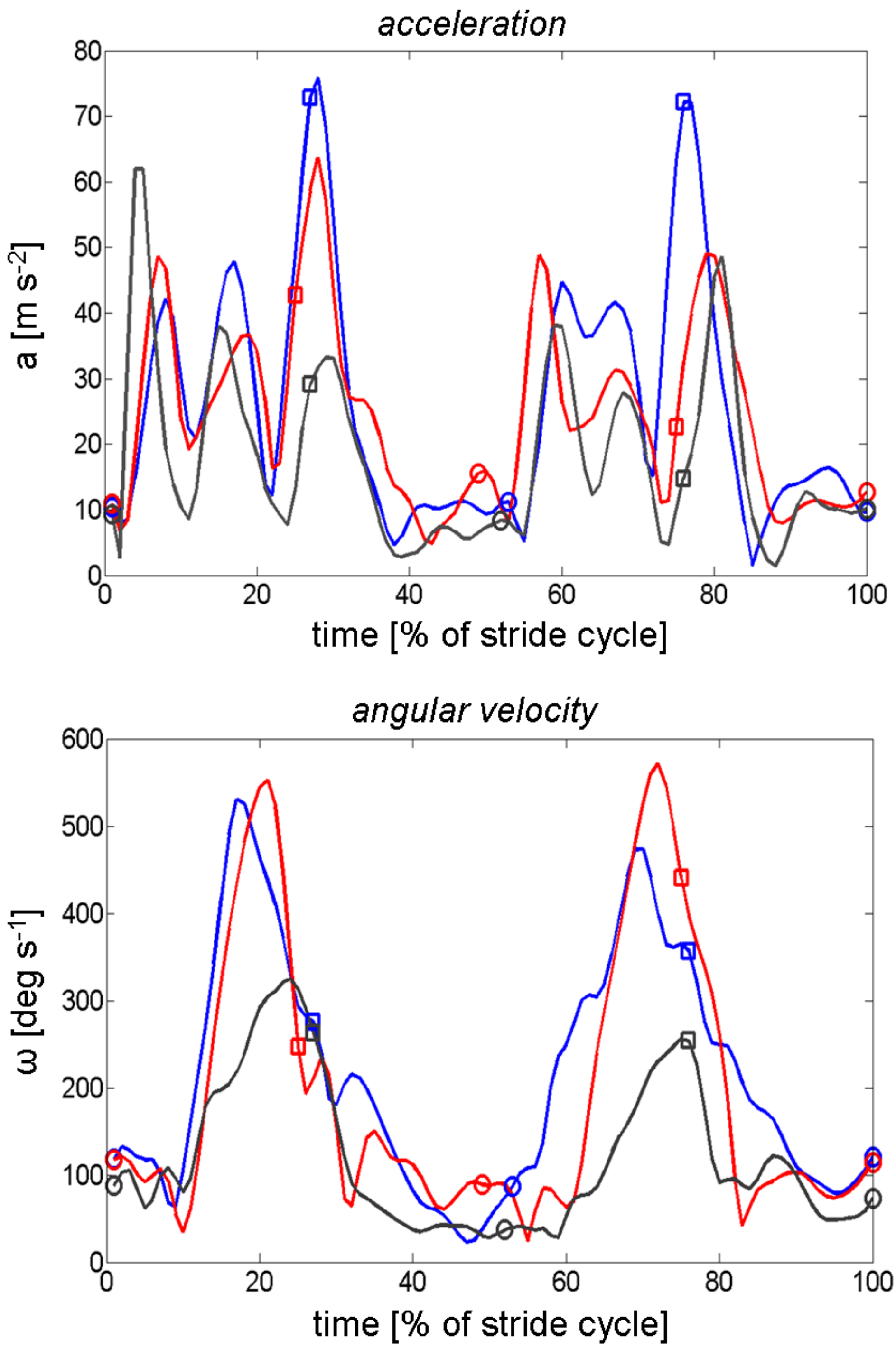


Figure 2 (3.4): Acceleration and angular velocity signals for three randomly selected subjects of group A, with reference to a randomly chosen stride cycle. Stride time is expressed in percentage of the total cycle duration. FS and FO events are indicated respectively with a circle and a square symbol.

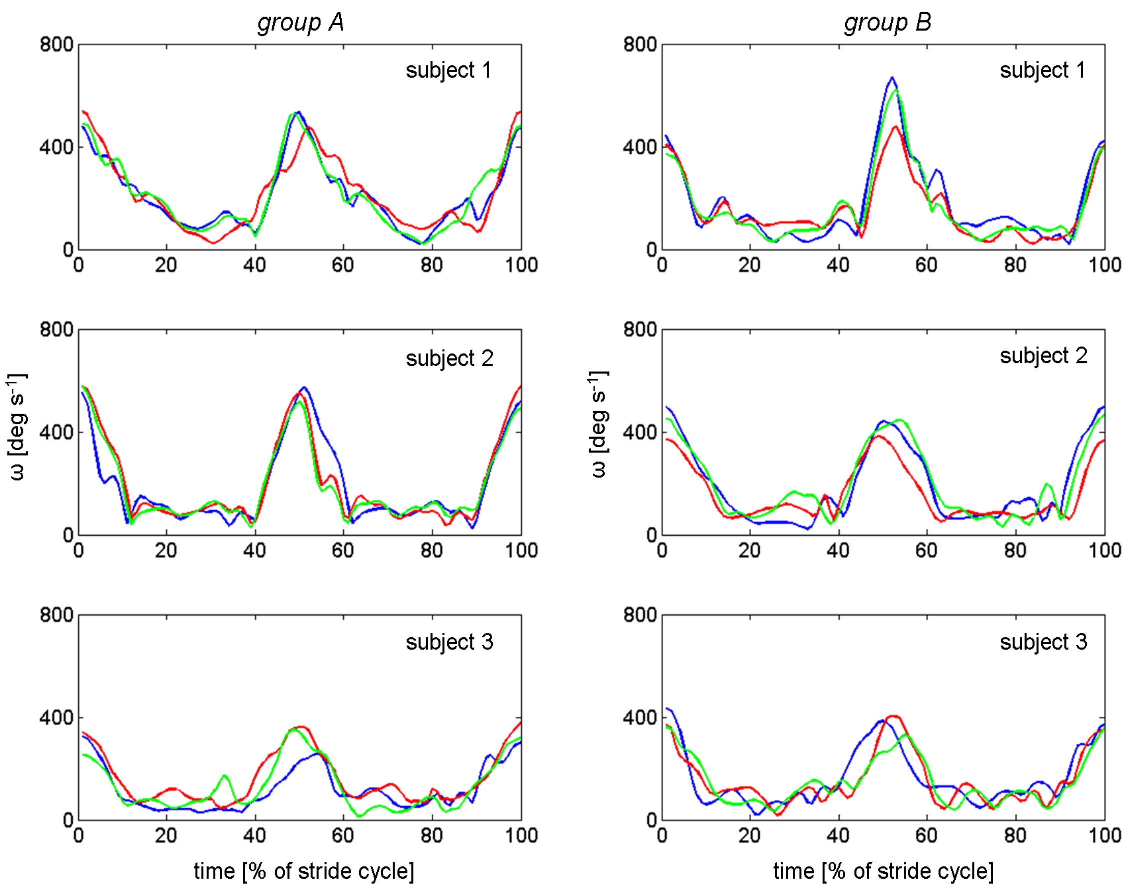


Figure 3 (3.4): ω for three subjects of group A (left panel) and of group B (right panel), with reference to three randomly chosen stride cycles (blue, red and green curves). ω_{peak} was used to segment stride periods for both group A and B. Each period is expressed in percentage of the total cycle duration.

FS and FO instants were identified from positive and negative peaks on the $\ddot{\omega}$ waveform (Fig. 4). These peaks were found to be consistently synchronized with FS and FO across steps, trials, subjects and groups.

No repeatable and consistent features, adequate for automatic detection of FS and FO related events, were identified on the two derivatives of a, neither in group A nor in group B.

Repeated-measures ANOVA performed on the estimated parameters $\tilde{d}_{stance}$ and $\tilde{d}_{stride}$ showed that no statistical differences across subjects (p > 0.05) and trial (p > 0.05) was present in both groups A and B. Conversely, as expected, statistical difference were found between groups (p < 0.05) for both $\tilde{d}_{stance}$ and

$\tilde{d}_{\text{stride}}$. Tab. 2 reports the mean ± one standard deviation values of $\tilde{d}_{\text{stance}}$ and $\tilde{d}_{\text{stride}}$ for both group A and group B.

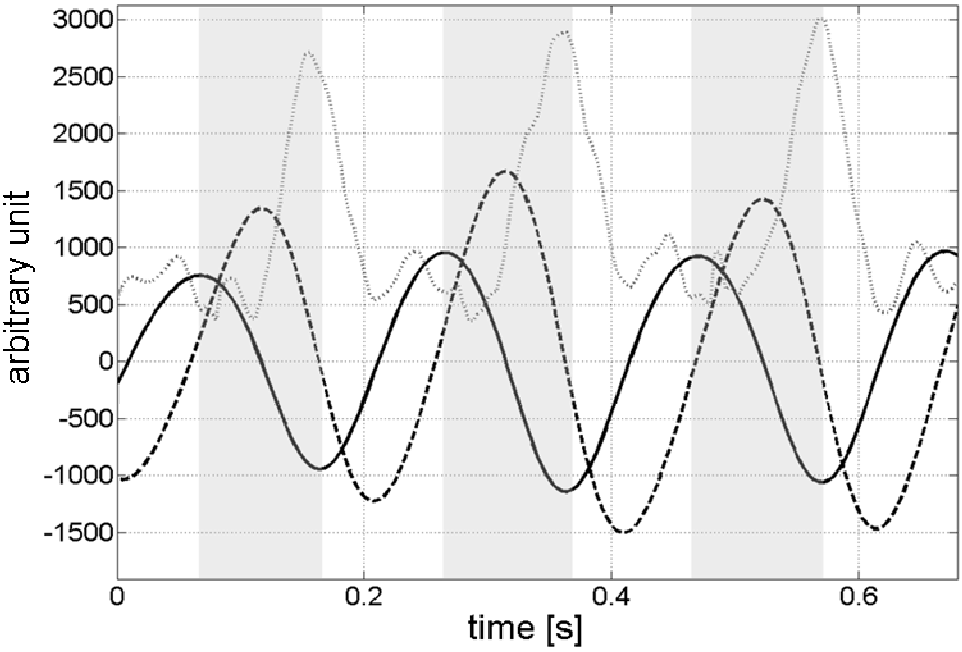
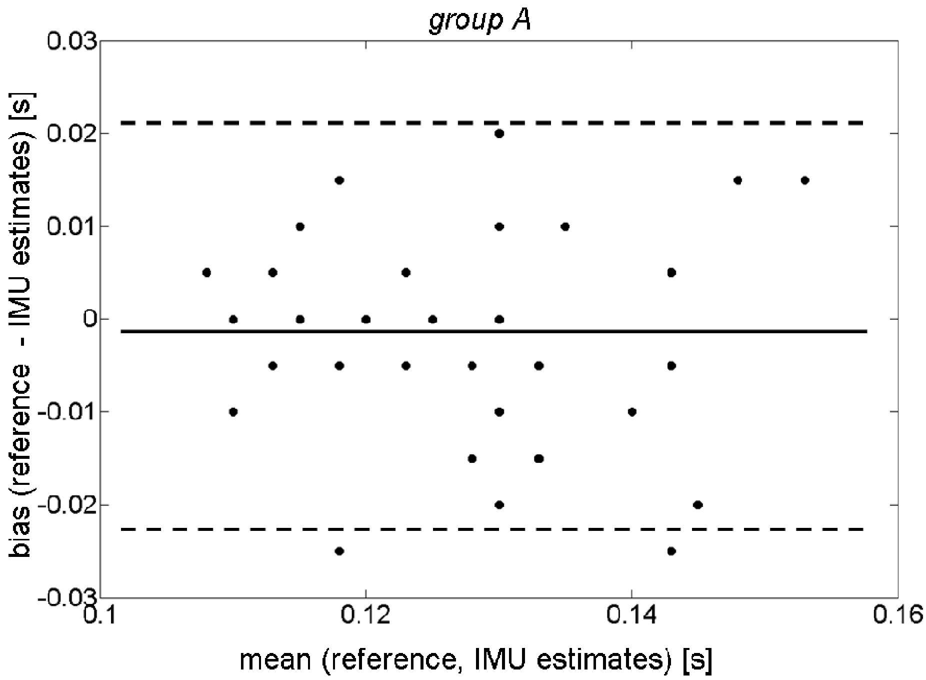


Figure 4 (3.4): ω (grey dotted line), $\dot{\omega}$ (dashed line) and $\ddot{\omega}$ (solid line) with reference to a randomly chosen subject of group A. Grey sections represent three consecutive stance phases.

	GROUP A		*GROUP B*	
	mean	**SD**	**mean**	**SD**
$\tilde{d}_{\text{stance}}$	0.125	± 0.015	0.105	± 0.010
$\tilde{d}_{\text{stride}}$	0.495	± 0.040	0.455	± 0.015

Table 2 (3.4): Mean ± 1SD of $\tilde{d}_{\text{stance}}$ and $\tilde{d}_{\text{stride}}$ across of all the subjects, trials and steps is provided for both group A and group B.

Bland and Altman plots illustrating the results about the accuracy of the stance duration estimates ($\tilde{d}_{\text{stance}}$) are shown in Fig. 5 for both group A (upper panel) and group B (lower panel). The mean of the absolute bias between the reference and the IMU estimates was 0.005 for both group A and B, as well as for both $\tilde{d}_{\text{stance}}$ and $\tilde{d}_{\text{stride}}$. The 95% of the limits of agreements was lower than 0.02 s for both groups. Symmetry of the bias limits of agreement (LOA) interval was observed, thus demonstrating that no systematic differences were present between reference and IMU estimates. No linear trend between the amount of bias and the measured values was found (average correlation = 0.02).



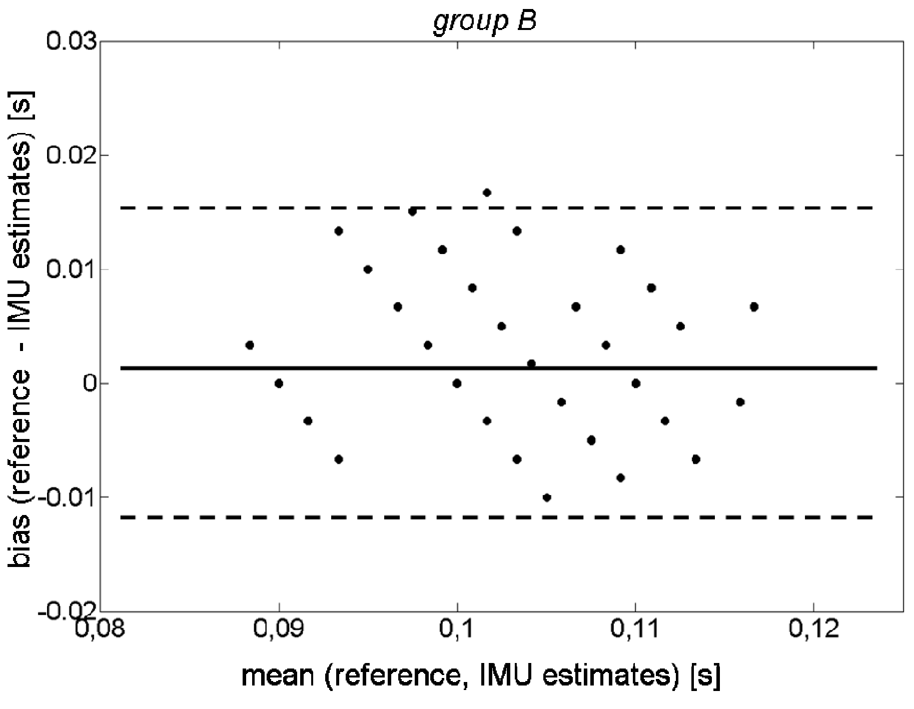


Figure 5 (3.4): Bland and Altman plots for group A and group B, corrected for the effect of repeated measurement error, representing comparisons between reference stance times and those estimated with IMU. Mean bias (solid line) and random error lines representing 95% limits of agreement (dashed lines) are included.

No differences were found across groups ($p > 0.05$), subjects ($p > 0.05$), and trials ($p > 0.05$) for each parameter bias.

3.4.4 Discussion

The feasibility of using a trunk-mounted IMU for estimating the stance duration during sprint running was assessed, and the accuracy of the IMU estimates quantified against reference measurements. The results show that neither the magnitude of the measured acceleration, nor its first and second derivatives provide any consistent and well identifiable feature correlated with FS and FO events. Conversely, the magnitude of the angular velocity vector, as well as its wavelet-mediated second derivative, are characterised by repeatable and consistent events correlated to stride and stance duration, respectively.

The identification on the acceleration time-series of features adequate for automatic detection and related to the FS and FO events was thwarted mainly by the low signal to noise ratio. The accelerometer signals, in fact, proved to be more subjected to the movement of the soft tissue masses relative to the underlying bones, with respect to the angular velocity measurements. Although each foot-ground impact was visible on the time history of the acceleration magnitude, neither FS nor FO events occurred at consistent maxima, minima, or even slope across different subjects. This seems to be in contrast with previous studies using trunk-mounted accelerometers during distance running. Wixted et al. (2010) reported, on a qualitative level, that the vertical, medio-lateral and antero-posterior acceleration curves of two professional middle-distance athletes have significant features that can be used in analysis of athletes' running style. Similarly, Auvinet et al. (2002) qualitatively showed a substantial consistency among the cranio-caudal, medio-lateral, and antero-posterior acceleration signals of seven professional middle-distance runners. However, neither study provides validation about the features consistency or about FS and FO event detection. In addition, each component of the accelerometer is subjected to the variability entailed in the sensor positioning. Using the magnitude of the acceleration vector means that the user can fix the sensor unit in any orientation, thus requiring no anatomical calibration or aligning protocols. This could be crucial in sports contexts, where the acquisition procedures must not interfere with the training session schedule. Finally, as distance running is characterised by a lower explosiveness with respect to sprint running, the amount of IMU oscillations relative to the underlying skeleton are lower, and the signal to noise ratio reasonably higher. This consideration makes the above mentioned study conclusions hardly applicable to sprint running.

The magnitude of the angular velocity signal presented a repeatable peak that proved to be consistent across groups, subject, and trials. As reported in the literature, this peak is associated to a trunk rotation about its longitudinal axis (Schache, et al., 2002). Such rotation occurs during walking as well as during running, and it is proved to be repeatable and to increase its amplitude with speed (Saunders, et al., 2005). Our results confirm the presence of this rotation during sprint running and prove that it is consistent among subjects with different expertise. Although this event can be used to identify the stride duration, it does not provide any information about the FS instant of time, thus about the stance duration.

Such parameter was consistently identified only when using a wavelet-mediated differentiation of ω that allowed for the identification on the $\ddot{\omega}$ waveform of positive and negative peaks associated to the FS and FO events. From a mechanical perspective, the local maxima associated to the FS is related to the maximal variation of the angular acceleration of the trunk, while the local minima associated to the FO is related to the trunk rotation about its longitudinal axis. Both features proved to be repeatable and consistent among all subjects and trials.

The absolute bias between the IMU estimates and the reference measurements was lower than 0.01 s for both $\tilde{d}_{stride}$ (< 2% stride duration) and $\tilde{d}_{stance}$ (< 8% stance duration). These results are consistent with those obtained by Purcell et al. (2006) as concerns the stance duration during sprint running using a shank-mounted accelerometer.

It has to be noted that the mean stance durations for both group A and B were higher than 0.1 s and that the mean difference of d_{stance} between the two groups of athletes involved in the present study was about 0.02 s, thus higher with respect to such bias. An error lower than 0.01 s appears to be acceptable also when considering the stance profile over time and during the whole race. Fig. 6 shows such profile for both the stance and the flight phases as obtained from a high-speed camera positioned behind the start line during one 80 m runs of an elite athlete of group B. The stance duration decreases of about 0.07 s during the acceleration phase of the run (0 to about 4 s). During this phase, the mean difference between one stance time and the following is 0.007 s, i.e. higher than the above mentioned mean bias between the IMU and the reference estimates (0.005 s). Furthermore, as such bias is of the same order than the temporal resolution of the IMU, it can be speculated that increasing that resolution may improve the final results.

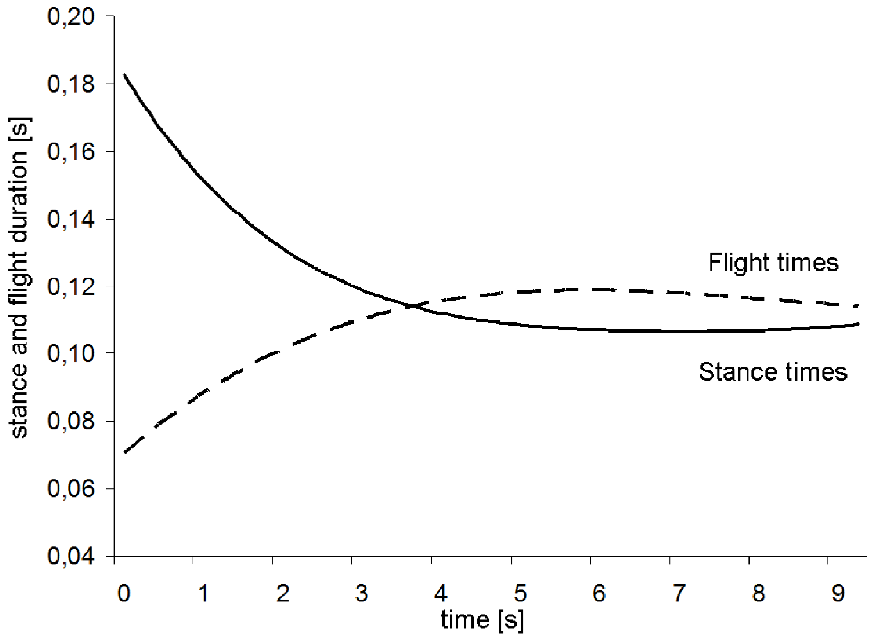


Figure 6 (3.4): Stance (solid line) and flight (dashed line) duration time-curve during a 80 m sprint run of an elite sprinter of group B.

In conclusion, stride and stance durations were estimated by using an IMU positioned on the lower-back trunk during the maximal speed phase of sprint running. In contrast to what expected, no consistent features were identified on the acceleration signal and on its first and second derivatives. Conversely, wavelet-mediated double differentiation of the angular velocity signal allowed for the identification of consistent and repeatable events correlated with FS and FO occurrences. Information about stance and stride durations could be particularly helpful for track and field trainers, being one of the key elements in sprint running performances. In order to provide coaches with an instrument that can be reliably and automatically used in the field, future works will concern the validation of the method on different phases of the sprint run.

3.5 LOW RESOLUTION APPROACH: DISCUSSION

The purpose of the Low Resolution Approach section was to develop methods for the assessment of performance-related biomechanical variables in sports applications. To this aim, the use of wearable inertial sensors for in-field performance evaluation of sprint running was discussed.

Results confirmed the agreement between the estimated quantities (instantaneous horizontal velocity, center of mass displacement, stride and stance durations, trunk inclination and angular velocity), and the reference data provided by stereophotogrammetry, force plates or high–speed cameras.

Two major sources of error proved to be crucial in the estimation of the above mentioned parameters: the movement of soft tissue masses relative to the underlying skeleton (soft tissue artefact) and the unstable bias characterising the sensors signals. The latter, in particular, rapidly accumulates over time when numerical integration of the measured signals is performed (for instance, to estimate instantaneous velocity or displacement from the acceleration, or angular displacement from the angular velocity). The explosiveness and high acceleration generated during sprint running and, in general, during sports activities emphasise the contribution of both sources of error.

The use of memory foam materials and elastic belts appears to be effective in limiting the soft tissues movement, while reducing the integration interval and exploring boundary conditions proved to cyclically correct the errors yielded by the unstable bias of the signal.

During sprint running, the acceleration signal was found to be more influenced by the soft tissue artefact with respect to the angular velocity signal, as attested by the impossibility to identify in the former repeatable and consistent features. This can be explained by considering that the soft tissue masses tend to move mainly along cranio-caudal and medio-lateral direction. Skin-mounted sensors measuring linear quantities seem, therefore, more sensitive to such artefact with respect to sensor measuring angular quantities.

It is worthwhile to underline that each new proposed method needs to be validated before being considered reliable and accurate. To this aim, the use of reference technologies characterised by higher accuracy with respect to the proposed method is necessary. As largely emphasised, traditional motion analysis instrumentations are limited in terms of portability and acquisition volume. Therefore, to perform in-field validations of methods involving inertial sensors, the currently available and used technologies are essentially photocells, laser-gun, in-shoe pressure sensors and high-speed cameras. Depending on the variable of interest, however, the accuracy of these technologies could be lower with respect to the accuracy of stereophotogrammetry or force plates, and sometimes could be inadequate. As a result, a step-by-step approach including in-lab validation could be, in some cases, mandatory. In particular, when the considered motor act could hardly be repeatable in a laboratory environment, the choice of analysing an equivalent motor paradigm could be a valid alternative. Such motor paradigm would be characterised by the same peak-to-peak values and rate of variation of acceleration than the motor act of interest (or angular velocity, or force depending on the aim of the study), with the advantage to be suitable for in-lab analysis.

That being said, inertial measurement units proved to be a valid alternative to traditional laboratory-based instrumentations when performing in-field evaluation of sprint running. Guidelines about the unit fixation appear particularly helpful and suitable for sports application. The development of algorithms that can minimise the detrimental effects of the above mentioned sources of error, while transforming this data into reliable parameters suitable for use by coaches and athletes continues to be a challenging exercise.

CHAPTER 4

HIGH RESOLUTION APPROACH

- *E. Bergamini, H. Pillet, J. Hausselle, P. Thoreux, S. Guerard, V. Camomilla, A. Cappozzo, W. Skalli – "Tibio-femoral joint constraints for bone pose estimation during movement using multi-body optimization". Gait & Posture, 2011. In press.*

- *E. Bergamini, H. Pillet, J. Hausselle, P. Thoreux, V. Camomilla, A. Cappozzo, W. Skalli – "Distance variation between origins and insertions of the knee ligaments during flexion-extension", 2nd National Bioengineering Conference, July 2010, Torino (Italy).*
- *H. Pillet, E. Bergamini, J. Hausselle, P. Thoreux, A. Cappozzo, W. Skalli – "Repeatability of a method to evaluate knee ligaments lengths during flexion-extension", 10th Annual Meeting of the International Society for Computer Assisted Orthopaedic Surgery, June 2010, Paris (France).*
- *P. Thoreux, H. Pillet, E. Bergamini, J. Hausselle, A. Cappozzo, W. Skalli – "Répétabilité d'une méthode d'évaluation de la variation de longueur des ligaments du genou au cours d'un mouvement de flexion-extension", 85th Meeting of the Société Française de Chirurgie Orthopédique et Traumatologique, November 2010, Paris (France).*

ABSTRACT

Musculoskeletal injuries are an undeniable facet of sports life of each athlete. A clear understanding of the mechanisms of musculoskeletal injury can aid in the effective diagnosis and treatment of the athlete.

Sprint running is characterised by the application of relatively high impact forces and large force impulses. Epidemiological studies of sprint running injuries found the knee to be the most frequent site of injuries. In this respect, forces and force-related factors appear to be the prime agents that determine the likelihood and severity of such injuries.

The estimate of external and internal forces in sports biomechanics often requires the use of inverse dynamics methods. The accuracy of these methods is affected by several sources of error. Among them, the motion of soft tissues relative to the underlying bones (soft tissue artefact – STA) is considered as the most important, particularly in sports applications where high acceleration and forces involved may be cause of even larger soft tissue motion.

In order to compensate for STA, several analytical techniques have been proposed. The so called multi-body optimization (MBO), in particular, aims at optimally estimating the location in space of a chain of bones interconnected in joints embedding specified constraints. To further improve the quality of constraint-based models of the joints, the definition of subject-specific, non-rigid constraints is desirable.

In this framework, the subject-specific plausible range of values of the lengths of the four major knee ligaments have been assessed during continuous loaded knee flexion. The inaccuracies associated with the estimation of both the subject-specific bone-models and ligament attachment locations have also been considered. Such plausible range of values are aimed at being incorporated, as non-rigid constraints, in a subject-specific kinematic model of the human knee which, hopefully, will allow for the improvement of the inverse dynamics method accuracy.

KEYWORDS: Injury risk factors; Running; Joint dynamics; Soft tissue artefact; Knee joint constraints.

4.1 INTRODUCTION

4.1.1 Sport biomechanics and injury prevention

This section was written on the basis of a book chapter by Zernicke, R.F., & Whiting, W.C. (2000). Mechanisms of musculoskeletal injury in biomechanics in sport: performance enhancement and injury prevention. In: Zatsiorsky, V.M. (Ed.), IOC Medical Commission, International Federation of Sports. The authors' work is gratefully acknowledged.

Injury is an undeniable facet of athletic life. Individuals across the spectrum of competitive levels, from recreational athletes to Olympic competitors, are subject to the limitations and disappointments imposed by musculoskeletal injury. Elite athletes, in particular, are highly susceptible to the physical, emotional, and economic costs that injury exacts. Though these costs are inconsequential for minor injuries, they can be overwhelming in cases of devastating injury that make it impossible, either temporarily or permanently, for an athlete to train and compete.

Lessening the chance of injury and facilitating the treatment and recovery of injured athletes requires an interdisciplinary approach that includes consideration of the anatomical, physiological, medical, kinesiological, psychological, and mechanical aspects of athletes and their competitive environments. Unfortunately, injury-related problems of athletes are not addressed optimally by each discipline acting in isolation, especially for elite athletes whose performances are too often considered more important than their health. An interdisciplinary team, including medical professionals, physical therapists, athletic trainers, sport scientists, psychologists, and coaches, is essential to bring a specific perspective to the challenges of dealing with injuries to athletes. In addition to these individual perspectives, musculoskeletal injury can be viewed from many different general perspectives, including historical, epidemiological, economic, psychological and biomechanical.

Of all scientific disciplines, physics and its sub-discipline, mechanics, are undeniably most central to the study of injury, since most injuries have mechanically related causes. The nature of an injury and its treatment rests on the effective integration of biological and mechanical knowledge. Interdisciplinary biomechanics (i.e. the application of mechanical principles to biological problems) is ideally positioned to provide this integrated perspective. In the case of musculoskeletal injury to athletes, the problems include those related to the prevention, diagnosis, and treatment of injuries so as to minimize their incidence and severity. The goal is to mitigate the negative effects of injury and allow athletes at all levels to maximise their performance potential.

As above mentioned, a comprehensive analysis of a movement pattern or skill requires not only the assessment of movement from a descriptive point of view (kinematics), but also knowledge of the causes that produce the movement (dynamics or kinetics). The word "cause" in mechanics, is always associated with the concept of force, defined as the mechanical action applied to a body that tends to produce acceleration.

Body movements are produced and controlled by forces acting both within the body (e.g. muscle forces) and upon the body from external sources (e.g. gravity, impact). These forces, in addition to modulating movement patterns, also act on tissues in the body. Under most conditions, these forces are well tolerated by the body's tissues. Indeed, force is central to optimal growth and development of tissues, especially those such as bone that provide structural support. But when forces exceed a tissue's ability to withstand the load, injury occurs. Forces and force-related factors (e.g. energy), therefore, are the most fundamental element in injuries and the prime agents that determine the likelihood and severity of a specific injury (Zernicke & Whiting, 2000).

Forces pertinent to injury analysis include those that act in or upon the human body, and include gravity, impact (e.g. a runner's foot contact with the ground), musculo-tendinous forces, ligamentous forces that stabilize joints, and forces applied to bones. For all injury-related forces, different factors combine to determine the nature and severity of the injury (Zernicke & Whiting, 2000): the magnitude, application point, line of action and duration of the applied force, as well as its frequency and rate of variability.

During athletic activities, body tissues are continuously subjected to forces that vary in magnitude, location, direction, duration, frequency, variability and rate. In most cases, the body tolerates these forces well, but when loads exceed the normal physiological range, a not uncommon occurrence in elite athletes, tissues can experience overload and sustain injury. Injuries resulting from a single or a few overload episodes are called *acute* injuries. These injuries often happen with a violent impact, such as when an object such as a ball contacts an athlete, or when an athlete impact on the ground, like a runner at the foot-ground contact followiong the flight phase, or a gymnast landing after a somersault. In general, as the area of force application increases, thus decreasing the pressure, the likelihood of injury decreases. When repeated loading leads to tissue damage, the injury is called an *overuse* or *chronic* injury. Calcaneal (Achilles) tendonitis in a long-distance runner provides one example of a chronic, overuse injury (Zernicke & Whiting, 2000).

Sprint running injuries: epidemiology

Epidemiological studies of running injuries found the knee to be the most frequent site of injuries (Clement, Taunton, Smart, & McNicol, 1981; Maughan & Miller, 1983; Brunet, Cook, Brinker, & Dickinson, 1990) with chondromalacia patella, pain on the undersurface of the patella, as one of the most frequent knee injuries. Tab. 1 summarises the results from four epidemiological studies (James, Bates, & Osternig, 1978; Clement, Taunton, Smart, & McNicol, 1981; Bennell & Crossley, 1996; Ballas, Tytko, & Cookson, 1997) that attempted to identify the source of lower-extremity injuries in running. It should be noted that the methods used to collect injury statistics and the specific population sampled can have a major effect on results, and such factors may account for some of the large differences seen in the reported studies.

There is in general a paucity of scientific studies showing definitive relationships between either anatomical factors and injury, or biomechanical measures and injury. Nevertheless, kinetic factors that are thought to cause knee injuries in runners include ground reaction and knee joint forces and moments (Ferber, Kingma, Bruijn, & van Dieen, 2002).

Grimston et al. (1994) and Ferber et al. (2002) found greater ground reaction forces in runners with stress fractures than in healthy runners. The vertical impulse and propulsive forces, moreover, were found to be greater in injured versus non-injured runners, but maximum vertical impact forces and loading rates were lower in the injured groups (Messier & Pittala, 1988; Messier, Davis, Curl, Lowery, & Pack, 1991; Messier et al., 1995; McCrory et al., 1999; Duffey, Martin, Cannon, Craven, & Messier, 2000). Flynn and Soutas-Little (1995) estimated that patellofemoral compressive forces ranged between 4.3 and 6.9 BW (Body Weight). Studies on lower extremity loads and overuse injury have shown that impact forces at commonly injured sites are extremely high (Scott & Winter, 1990), and previously injured runners have higher lower extremity loads (Milner, Davis, et al., 2006; Milner, Ferber, et al., 2006) and greater hip joint range of motion than non-injured runners (Noehren, Davis, & Hamill, 2007). Hreljac (2004) suggests that the repeated application of relatively high-impact forces, as happens in sprint running, without sufficient time between them to allow positive remodelling, ultimately results in an overuse injury.

	Study				
	James *et al.* (1978)	Clement *et al.* (1981)	Ballas *et al.* (1997)	Bennell and Crossley (1996)	Bennell and Crossley (1996)
Type of subjects	Runners	Runners	Runners	Runners	Sprinters
No. of injuries	180	1650	860	39	19
Type of data	Clinic	Clinic	Clinic	Interview	Interview
Site of injury (%)					
Knee pain	29.0	25.8	13.8	15.0	14.0
Shin splints–tibial stress syndrome	13.0	13.2	7.8	13.6	5.0
Achilles tendinitis	11.1	6.0	2.2		
Plantar fasciitis	7.0	4.7	4.0		
Ankle/foot tendinitis				13.9	6.0
Stress fracture	6.0	5.8	9.3	25.1	18.0
Tibial stress fracture		2.6			
Metatarsal stress fracture		3.2			
Iliotibial tract tendinitis	5.0	4.3	3.8		
Patellar tendinitis		4.5	2.2		
Hamstring strain			5.2	4.3	38.0
Adductor strain			6.0		
Ankle lateral ligament sprain			4.9	8.9	3.0
Others			9.4		

Table 1 (4.1): Common injury sites in running (from Zernicke & Whiting, 2000).

Although these data do not provide unequivocal support, it appears that high forces applied to the lower extremity tissues during running are associated with running-related injuries and therefore may be mechanisms of knee injuries. An accurate estimate of such forces appears to be crucial in improving injury prevention strategies and athlete effective treatment.

4.1.2 Joint dynamics estimation

Different techniques can be used to determine joint dynamics variables (forces, moments and powers):

- Dynamometry: measurement of all or part of the scalar quantities that describe a force system. Force platforms, based on strain-gauge or piezoelectric force transducers are among the most used dynamometer in human movement analysis.
- Inverse dynamics: estimation of the forces based on their observed effects on the kinematics of the athlete's body, i.e. on the time histories of the position and orientation of the athlete's body segments. Kinematic data are collected by means of stereophotogrammetric motion capture systems. It has to be noted that this approach allows for the determination of the resultant rather than individual forces acting on the selected body part.

Although inverse dynamics is typically less accurate than direct measurement, in some cases it is mandatory because the letter technique is not applicable. When dealing with joint dynamics, for instance, it is often impossible to directly measure internal forces such as those transmitted by muscles, tendons or bones. This, indeed, would require the surgical implantation of a dynamometer in the above mentioned intra or periarticular structures. In sports biomechanics, moreover, it is even very difficult to measure external forces: dynamometers can hardly be placed in fields, tracks, boards, gymnasium floors or equipment used for sport competitions. On the other hand, as widely discussed in previous sections, the majority of sports motor tasks can not be easily repeated in a motion analysis laboratory, not mentioning athletes' performances during competitions.

The accuracy of inverse dynamics is limited by several different sources of error, including the simplifying assumptions used to define the mathematical model of the human body. Among them, errors affecting the estimation of body segment inertia parameters and those derived from the movement of soft tissues (essentially muscles, fat tissue and skin) relative to the underlying bones are the most critical in sports biomechanics.

While methods for accurately estimate inertial parameters have been proposed and showed to be effective and relatively easy to be applied (de Leva, 1996), the errors associated to the soft tissue artefacts are much more difficult to minimise and compensate for (Cappozzo, 1991; Leardini, Chiari, Della Croce, & Cappozzo, 2005; de Leva & Cappozzo, 2006; Peters, Galna, Sangeux, Morris, & Baker, 2010).

Different methods have been proposed to model the motion of the soft tissue masses and of the bones separately, in order to obtain accurate estimates of the resultant force applied on the body (Gruber, Ruder, Denoth, & Schneider, 1998; Pain & Challis, 2006). De Leva and Cappozzo (2006) reported a peak error value of -0.9 BW between measured and estimated vertical GRF of a volleyball player performing four consecutive counter movement jumps with free arms. Their results confirm that the difference between sports activities and daily motor tasks considerably amplify the magnitude of STA and of its consequence on the estimate of dynamic variables.

In order to compensate for STA, several analytical techniques have also been proposed (Leardini, Chiari, Della Croce, & Cappozzo, 2005). All of them exploit redundancy of measured information and embed a model of the artefact (Soderkvist & Wedin, 1993; Cappello, Cappozzo, LaPalombara, Lucchetti, & Leardini, 1997; Lucchetti, Cappozzo, Cappello, & Della Croce, 1998; Alexander & Andriacchi, 2001). These techniques are either used to optimally estimate the pose of one bony segment at a time (single-body optimization – SBO), and deal with all of its six degrees of freedom, or aim at optimally estimating the location in space of a chain of bones interconnected in joints embedding specified constraints (multi-body optimization – MBO; in the literature often referred to as global optimization). In the latter case, indeed, the available information about the predictable behaviour of the main human joints, which imposes constraints to the motion of the bones relative to each other, is aimed to be exploited. Recent studies argue that more advanced joint models could actually reduce the errors associated with skin marker-based measurement of joint (Andersen et al., 2009; Duprey, et al., 2010). In this respect, the definition of proper joint constraints is crucial. To this aim, the anatomy constitutes an ideal candidate to define subject-specific, realistic and non-rigid constraints for models of the joint of interest.

4.2 STUDY 4: TIBIO-FEMORAL JOINT CONSTRAINTS FOR BONE POSE ESTIMATION DURING MOVEMENT USING MULTI-BODY OPTIMIZATION

Abstract

When using skin markers and stereophotogrammetry for movement analysis, bone pose estimation may be performed using multi-body optimization with the intent of reducing the effect of soft tissue artefacts. When the joint of interest is the knee, improvement of this approach requires defining subject-specific relevant kinematic constraints. The aim of this work was to provide these constraints in the form of plausible values for the distances between origin and insertion of the main ligaments (ligament lengths), during loaded healthy knee flexion, taking into account the indeterminacies associated with landmark identification during anatomical calibration.

Ligament attachment sites were identified through virtual palpation on digital bone templates. Attachments sites were estimated for six knee specimens by matching the femur and tibia templates to low-dose stereoradiography images. Movement data were obtained using stereophotogrammetry and pin markers. Relevant ligament lengths for the anterior and posterior cruciate, lateral collateral, and deep and superficial bundles of the medial collateral ligaments (ACL, PCL, LCL, MCLdeep, MCLsup) were calculated. The effect of landmark identification variability was evaluated performing a Monte Carlo simulation on the coordinates of the origin-insertion centroids. The ACL and LCL lengths were found to decrease, and the MCLdeep length to increase significantly during flexion, while variations in PCL and MCLsup length was concealed by the experimental indeterminacy**.**

An analytical model is given that provides subject-specific plausible ligament length variations as functions of the knee flexion angle and that can be incorporated in a multi-body optimization procedure.

Keywords: Knee ligament length; Joint constraints; Soft tissue artefact; Global optimization; Biomechanics.

4.2.1 Introduction

Human movement analysis is largely based on the use of skin markers and stereophotogrammetry. Since markers are not rigidly associated with the underlying bone, bone pose reconstruction is affected by a soft tissue artefact (STA). The magnitude of this artefact and its disruptive consequences on the accuracy of the mechanical analysis of a motor act are well described in the literature (Peters, et al., 2010).

In order to compensate for STA, several techniques have been proposed. All of them exploit redundancy of measured information and embed a model of the artefact (Soderkvist & Wedin, 1993; Chèze, Fregly, & Dimnet, 1995; Alexander & Andriacchi, 2001). These techniques are either used to optimally estimate the pose of one bony segment at a time (single-body optimization – SBO), and deal with all of its six degrees of freedom, or aim at optimally estimating the location in space of a chain of bones interconnected in joints embedding specified constraints (multi-body optimization – MBO; often referred to as global optimization).

As opposed to the SBO, which deals only with the artefact deformation of the cluster of markers, the MBO compensates also for its overall displacement relative to the skeleton. In addition, the MBO may provide more realistic joint kinematics by, for instance, preventing bones from appearing to macroscopically pierce into each other. It should be noted however, that more realistic does not necessarily mean more accurate and that, when using the MBO, the joint degrees of freedom that are not embedded in the model are sacrificed.

Thus, MBO should not be expected to improve the estimate of joint kinematics (Andersen, Benoit, Damsgaard, Ramsey, & Rasmussen, 2010), but to provide an optimal reconstruction of the instantaneous location in space of the entire chain of bones being analyzed. Therefore, a more accurate kinetic analysis is made possible and the reconstruction of the movement of the soft tissue masses relative to the underlying bone may be attempted. The latter movement may be fed into a biodynamic model of the human body, the kinetic analysis of which would incorporate the relevant inertial effects that are known to significantly affect the forces involved in motor acts characterized by high accelerations (de Leva & Cappozzo, 2006). These considerations justify investing resources aimed at improving the MBO approach.

MBO has been performed mainly using joint constraints which prevent translation, such as spherical, revolute or universal couplings (Lu & O'Connor, 1999; Andersen, et al., 2009). The knee has also been modelled using a parallel mechanism incorporating articular surfaces, as sphere-on-plane contacts, and isometric pseudo ligamentous structures, and allowing for both translations and rotations (Duprey, et al., 2010).

To further improve the quality of these models, it is desirable to develop a subject-specific kinematic model of the human knee which, during movement, incorporates more realistic non-rigid constraints to the distances between the origin and insertion landmarks (for simplicity referred to as lengths) of the four major ligaments: anterior and posterior cruciate, and lateral and medial collateral ligaments (ACL, PCL, LCL, MCL). These constraints can be represented by the subject-specific plausible range of values of the ligament lengths for any given flexion angle of the knee. It should be noted that these ranges of values, in order to be incorporated in the above-mentioned knee model, must account for the indeterminacies associated with the estimation of the subject-specific digital models of the bones involved and of the location of the ligament origin and insertion landmarks (anatomical calibration).

Several authors have focused on ligament kinematic behaviour with the aim to support ligament injury treatment: ACL and PCL elongation was investigated *ex-vivo* (Grood, Hefzy, & Lindenfield, 1989; Hollis, Takai, Adams, Horibe, & Woo, 1991) and *in-vivo* (Li, DeFrate, Sun, & Gill, 2004; Hosseini, Gill, & Li, 2009; Yoo et al., 2010). Fewer studies have dealt with the MCL and LCL (Van de Velde et al., 2007). However, most of these studies provide ligament lengths only for a few knee flexion angles and do not display a general consensus about the amount and direction of ligament length variation during flexion. In addition, the errors associated with origin and insertion landmark identification have not been accounted for. Thus, no general boundary conditions, as required in constraint-based knee modelling, can be derived for the ligament lengths during continuous flexion-extension.

The purpose of this study was to determine the above-mentioned plausible values for the ligament lengths during loaded continuous healthy knee flexion. To this aim, *ex-vivo* tibio-femoral joints were used to avoid soft tissue artefacts. Three-dimensional digital models of tibias and femura were reconstructed and ligament attachment areas identified on them together with the associated intra- and inter-operator indeterminacies. The bone models were made to move *in-silico* using experimental data obtained on the joint specimens. Ligament lengths were thereafter estimated as a function of the joint flexion angle and submitted to statistical analysis in order to provide a knee joint model that could be embedded in the MBO procedure.

4.2.2 Materials and methods

Six knee specimens, consisting of femur, patella, fibula and tibia and intact joint passive structures, were harvested from subjects aged 75 to 96 years old and fresh frozen. They exhibited no advanced osteoarthritis or ligament laxity and presented no recurvatum.

The knee specimens were set in motion using a device described in Azmy et al. (Azmy, Guérard, Bonnet, Gabrielli, & Skalli, 2010). The femur was fixed to the experimental jig and the tibia caused to move by a servo-actuator that applied a force to the quadriceps tendon that simulated a contraction of this muscle group. A flexion resistive moment was applied to the knee through two cables fixed to the distal end of the tibia and passing through two pulleys coaxial with the centre of the femoral head and located laterally and medially to it. The force transmitted by these cables to the tibia was equal to 30 N and maintained bone coaptation.

Clusters made of three retro-reflective markers, with a minimal separation of 50 mm, were secured to the femur and to the tibia, each using two pins (Fig. 1a, 1b). The marker-cluster 3D positions and orientations were reconstructed during movement using a stereophotogrammetric system (Polaris, Nothern Digital Inc., Canada) at a rate of 60 samples per second.

3D digital template models of femur and tibia were obtained from CT-scan images. The perimeters of the areas of origin and insertion of the ACL, PCL, LCL, as well as of the deep and superficial bundles of the MCL (MCLdeep, MCLsup), were identified and traced on the bone templates using the mouse pointer. This procedure was performed seven times by one operator and once by three different operators for a total of ten virtual palpations. The professionals involved were orthopaedic surgeons and underwent a specific and meticulous training.

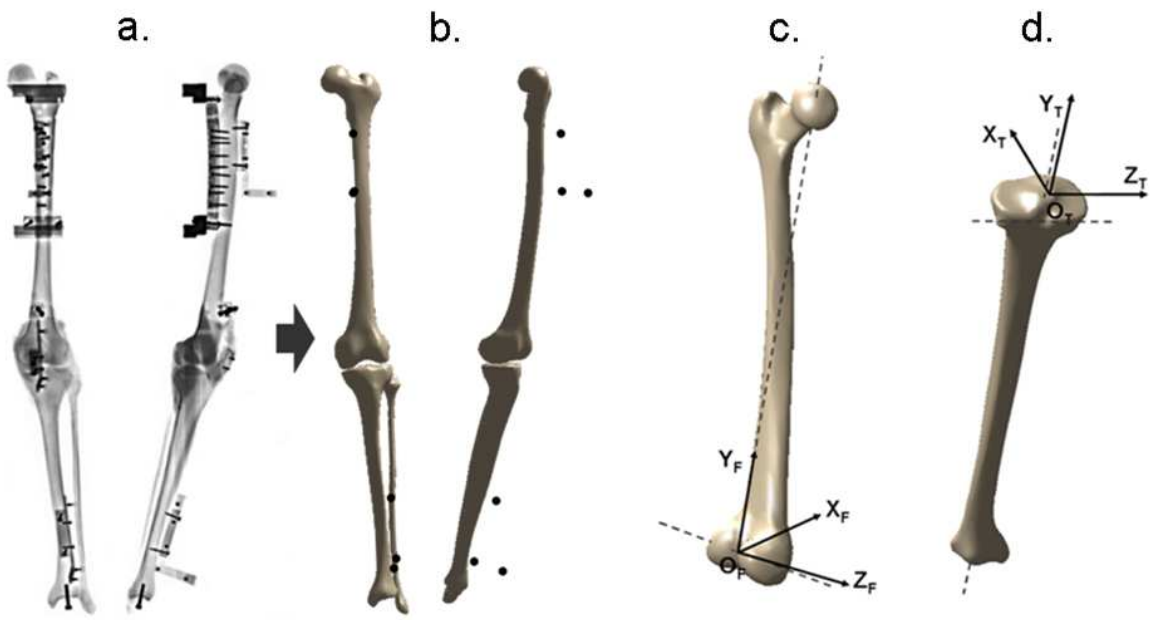


Figure 1 (4.2): (**a**) 2D images acquired with the stereoradiographic system. (**b**) Reconstruction of the femur and tibia 3D digital models together with the photogrammetric markers. (**c**) Femur system of reference [18]: O_F (origin): mid point of the segment joining the centers of the two condylar spheres, obtained by least squares approximation of the posterior portion of the medial and lateral epicondyles; Y_F: axis going from O_F to the centre of the femoral head; Z_F: projection onto the plane orthogonal to Y_F of the segment joining the centres of the two condylar spheres (dashed line); X_F: cross product between Y_F and Z_F. (**d**) Tibia system of reference [18]: O_T (origin): centroid of the tibial plateaux; Y_T: axis going from the centroid of the tibial pilon surface to the intersection between the principal inertial axis of the tibial diaphysis (dashed line) and the tibial plateaux surface; Z_T: projection onto the plane orthogonal to Y_T of the segment joining the most posterior points of the tibial plateaux (dashed line); X_T: cross product between Y_T and Z_T.

Two orthogonal digital radiographs of each knee specimen were simultaneously obtained using a low dosage X-ray system (EOS®, EOS-imaging, France) (Fig. 1a). The 3D bone-models were obtained through a reconstruction algorithm based on three steps: 1. identification and labelling of anatomical landmarks on the radiographic images in order to set a parametric simplified subject-specific model; 2. pre-morphing of the bone template to get an initial estimate of the bone; 3. iterative deformation of the latter estimate, based on parametric models and statistical inferences, until the best estimate of the subject-specific bone-model, carrying marks indicating the selected ligament attachment areas, was obtained (Fig. 1b) (Chaibi, et al., 2011). The root mean square discrepancy between a digital model of a femur or tibia, as obtained using this procedure, and the relevant CT-scan model, was assessed in a previous study and found to be on average less than 1 mm (Chaibi, et al., 2011). Each subject-specific bone-model was reconstructed twice by three operators.

Each specimen was subjected to six flexion-extension cycles during which marker trajectories were recorded.

Anatomical frames of femur and tibia (Fig. 1c, 1d) were defined as suggested in Schlatterer et al. (2009).

Using the marker coordinates in the EOS® frame, the subject-specific bone-models and the relevant anatomical axes were registered with respect to the movement data given in the stereophotogrammetric (global) frame. The knee joint kinematics was then estimated using the Cardan convention and the sequence Z_T, X_T, Y_T.

For each of the six knees, each knee-model reconstruction, each flexion-extension cycle, and each virtual palpation, the centroids of the attachment areas of each ligament were determined. The mean values and standard deviations (SDs) of these centroid coordinates, as represented in the global frame, were calculated. It must be noted that virtual palpation did not display any significant difference between inter and intra-operator variability.

The Euclidean distances between the mean origin and insertion centroids, obtained averaging the ten virtual palpations, were then computed during the knee movement (d_c) (Fig. 2).

A similarity analysis of d_c vs. flexion angle curves, as obtained from the six flexion-extension cycles, was performed (r = 0.98±0.01; RMSE = 0.15±0.01 mm). Results showed that no significant hysteresis occurred, thus, only mean curves were considered for further analysis.

At this point, for each knee the data set was made of six curves, one for each knee-model reconstruction.

In order to assess the impact of the inaccuracies associated with the subject-specific bone-model reconstruction procedure on the estimation of ligament length, a parametric statistical analysis was carried out by making reference to the values of d_c at full knee extension (d_e). To this purpose the intraclass correlation coefficient (ICC) (McGraw & Wong, 1996) was calculated and the Bland and Altman plots with correlation analysis of the bias between each couple of operators inspected (Bland & Altman, 2010).

The propagation to the d_c values of the errors associated with the variability of the virtual palpation was assessed using a Monte Carlo simulation. Origin and insertion points were randomly generated, using a normally distributed mislocation from the mean centroid. The ligament length between each possible pair of the generated origin and insertion points (100*100 pairs) was computed during knee motion (d_{MC}) (Fig.2).

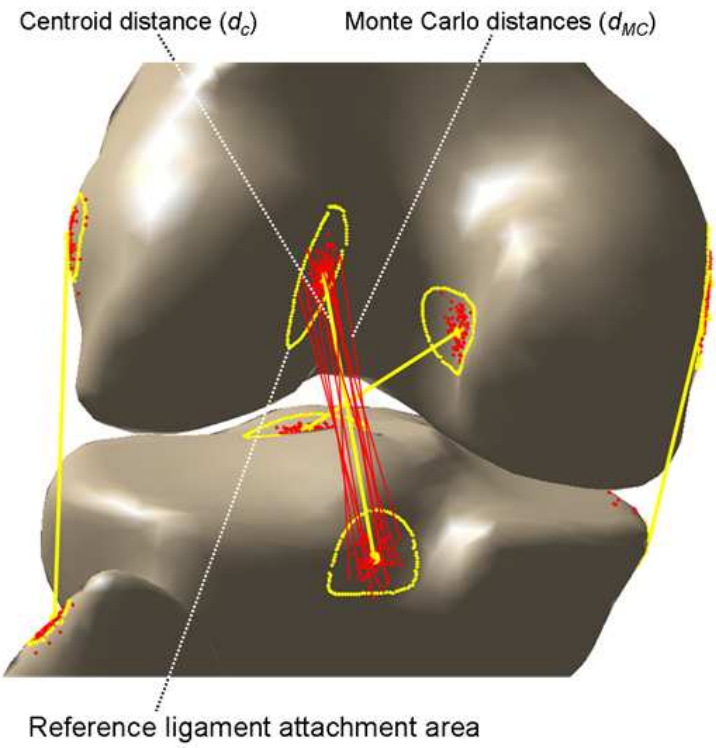


Figure 2 (4.2): Digital model of a randomly chosen knee specimen: the ligament attachment areas and the tibio-femoral distances between the centroids (d_c) (yellow lines) as well as between selected Monte Carlo pairs (d_{MC}) (red lines) are depicted.

For the sake of generalization, ligament length variations (Δd_{MC}) were then calculated relative to d_e and expressed as percentage of the latter value for each sampled knee flexion angle. The mean value and standard deviation of Δd_{MC} were then calculated for each Monte Carlo pair over the six knees and the six digital model reconstructions. To facilitate embedding this information in the knee kinematic model to be used in the MBO process, the mean of the curves Δd_{MC} vs flexion angle thus obtained, plus and minus one SD, were fitted with a polynomial regression function of the fifth order.

4.2.3 Results

Both the intra- and inter-operator variability displayed a SD lower than 2 mm for all ligaments with regard to the ligament origin and insertion coordinates on subject-specific knee models. Thus, this value was conservatively used as the SD of the normal distribution used in the Monte Carlo simulation.

With respect to the effect of the indeterminacies associated with the subject-specific bone-model reconstruction on the estimate of d_e, a high intra-operator (ICC = 0.936; Confidence Interval (CI) [0.894;0.961]) and inter-operator (ICC = 0.974; CI [0.960;0.983]) repeatability was found. Bland and Altman plots and correlation analysis on the residuals confirmed these results. The mean bias and the bias correlation computed between each pair of operators were low, with both CIs spanning the zero value, thus demonstrating that no systematic differences were present among the operators (bias = 0.11 mm; CI [-0.309;0.530]; corr = 0.13; CI [-0.116;0.373]). Given the resulting repeatability, multiple reconstructions from different operators were considered as repeated measures. The variability of d_e over the six observations and averaged over the six knees was measured by a SD equal to 2.3 mm for ACL, 1.9 mm for PCL, 2.4 mm for MCLdeep, 2.1 mm for MCLsup, and 1.7 mm for LCL.

In Fig. 3, the distance d_c is represented against the knee flexion angle with reference to a randomly chosen knee. The ACL, LCL, and MCLdeep displayed consistent trends in all knees, while the PCL and the MCLsup did not. In Fig. 3, the selected knee internal-external rotation and the ab-adduction angles are also depicted vs the flexion angle. The other knees displayed virtually identical curves.

The mean and SD curves of Δd_{MC} are shown in Fig. 4 for each ligament. The coefficients of the regression functions used to fit the displayed curves are reported in Tab. 1. The mean of the residual error norm between the fitting function and the original curve was lower than 0.7±0.4%.

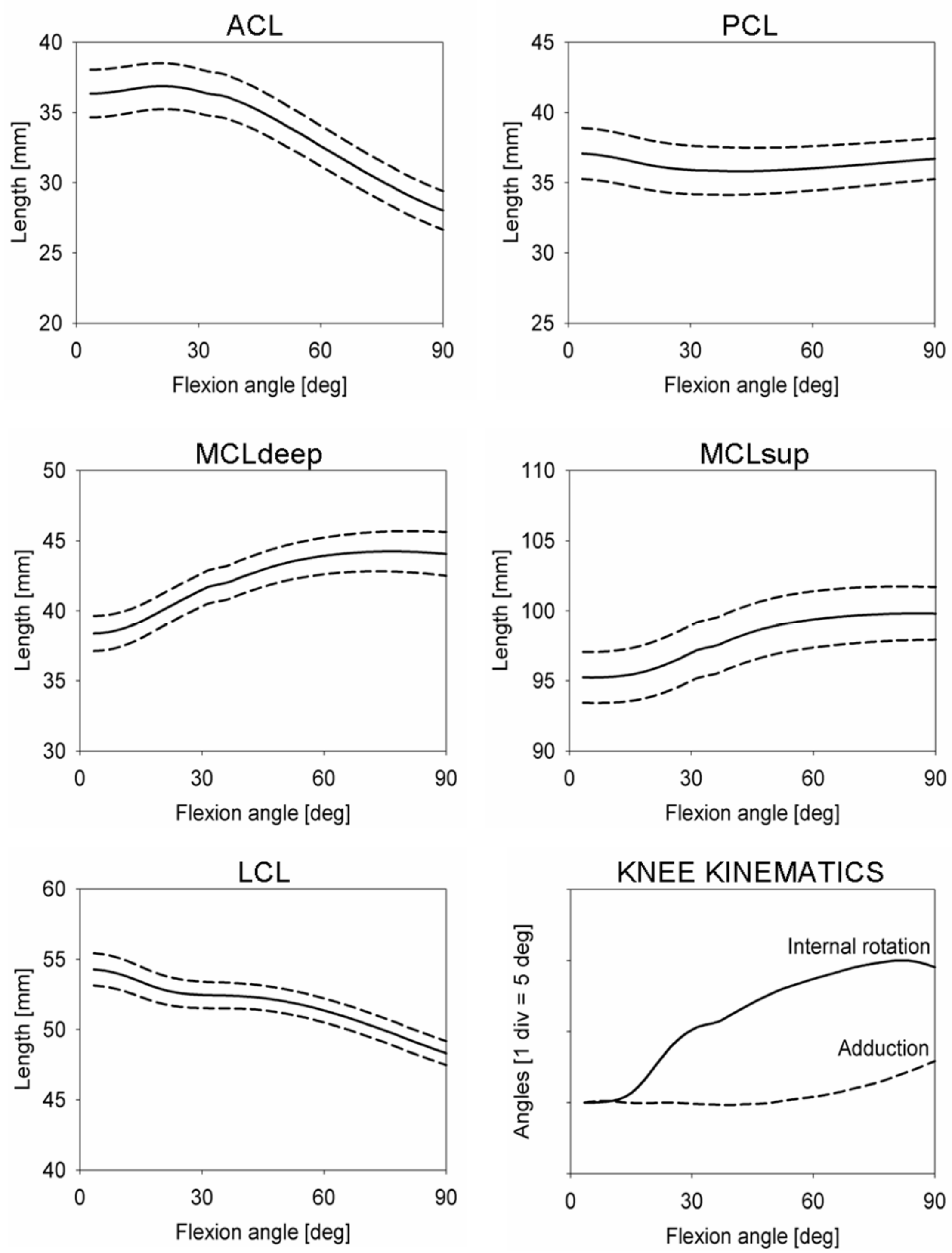


Figure 3 (4.2): For one randomly selected knee specimen, the ligament lengths are depicted as a function of the knee flexion angle: mean of three operators times two reconstructions ± 1SD. The kinematics of the same knee is also shown in the bottom right panel.

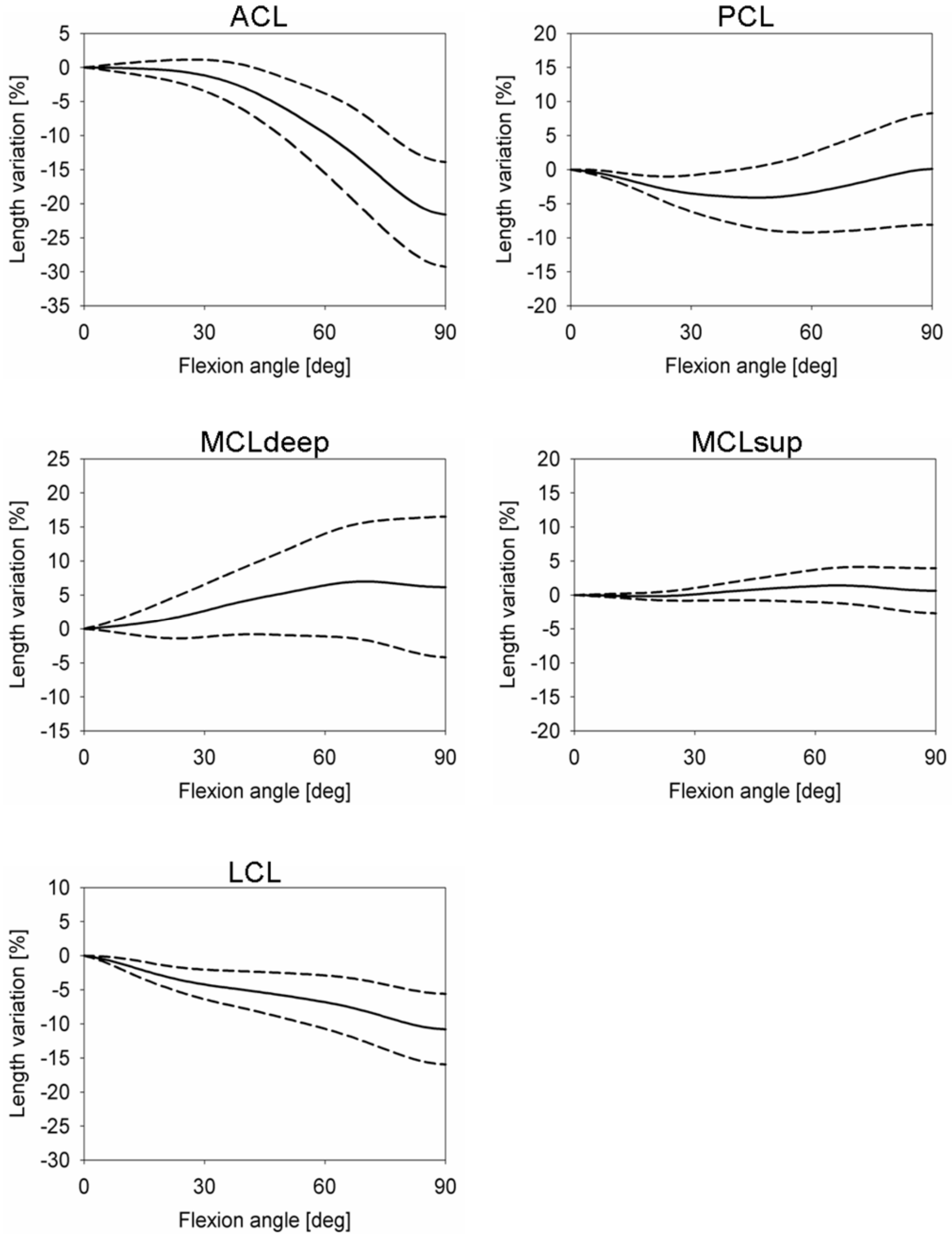


Figure 4 (4.2): Distance variation patterns (mean ± 1SD) vs the knee flexion angle as obtained through the Monte Carlo simulation for each ligament, Δd_{MC}. Each variation is expressed as a percentage of the distance at knee maximal extension.

		a_0	a_1	a_2	a_3	a_4	a_5
ACL	Mean	-4.3	-7.4	-3.9	-0.1	0.5	0.1
	+ 1SD	-0.4	-4.1	-3.5	-0.6	0.4	0.2
	- 1SD	-8.2	-10.7	-4.3	0.4	0.7	0.1
PCL	Mean	-4.1	0.1	2.4	0.0	-0.3	0.0
	+ 1SD	0.2	2.9	2.1	0.0	-0.2	-0.1
	- 1SD	-8.5	-2.7	2.7	-0.1	-0.4	0.1
MCLdeep	Mean	4.8	3.6	-0.8	-0.9	0.1	0.1
	+ 1SD	10.4	6.8	-0.7	-0.9	-0.1	0.1
	- 1SD	-0.9	0.4	-0.9	-1.0	0.2	0.1
MCLsup	Mean	0.8	1.3	-0.3	-0.7	0.0	0.1
	+ 1SD	2.4	2.6	-0.2	-0.9	0.0	0.1
	- 1SD	-0.8	0.0	-0.4	-0.5	0.1	0.1
LCL	Mean	-5.4	-1.9	-0.2	-1.2	0.1	0.2
	+ 1SD	-2.4	-0.4	-0.1	-1.1	0.0	0.2
	- 1SD	-8.5	-3.4	-0.3	-1.2	0.2	0.3

Table 1 (4.2): Coefficients of the fifth-order polynomial regression function used to fit the mean and SD distance curves obtained from the Monte Carlo pairs (see Fig. 4): d (dependent variable): percentage ligament length variation [%]; β (independent variable): knee flexion angle [deg]. In order to improve the fitting model accuracy and to reduce the influence of random errors on the regression coefficients, the independent variable β was standardized by computing its z-score, using its mean (μ) and the corresponding standard deviation (σ):

$z = \frac{(\beta - \mu)}{\sigma}$, where μ = 45 deg and σ = 26 deg.

The following regression equation was then used:

$$d = a_0 + a_1 \cdot z + a_2 \cdot z^2 + a_3 \cdot z^3 + a_4 \cdot z^4 + a_5 \cdot z^5 .$$

4.2.4 Discussion

To appropriately define knee joint constraints related to ligament geometry, the subject-specific plausible values of the lengths of the four major ligaments were assessed during continuous loaded knee flexion, while considering the inaccuracies associated with the estimation of both the subject-specific bone-models and ligament attachment locations.

The ACL and LCL were found to shorten during knee flexion reaching, on average, 22% and 11% of their length at full extension, respectively. This is in agreement with works reporting *in-vivo* behaviour of the ACL (Beynnon et al., 2001; Li, et al., 2004; Hosseini, et al., 2009; Yoo, et al., 2010), *ex-vivo* behaviour of the LCL (Harfe, Chuinard, Espinoza, Thomas, & Solomonow, 1998) and with a simulation study involving both ligaments (Amiri, Cooke, Kim, & Wyss, 2007).

PCL length increased or decreased depending on the selected origin and insertion location within the plausible area. This dependency has been already evidenced *ex-vivo* (Grood, et al., 1989; Ahmad et al., 2003) and by using a simulation approach (Amiri, et al., 2007). Conversely, *in-vivo* studies showed an elongation of the central bundle of the PCL (DeFrate, Gill, & Li, 2004), as well as of the antero-medial and postero-lateral ligament bundles (Li, et al., 2004). Since the patterns of ligament length variation depend on the kinematics of the joint (Woo, Debski, Withrow, & Janaushek, 1999; Amiri, et al., 2007), the lack of agreement between *in-vivo* and *ex-vivo*, as well as simulation results might be attributed to different amounts of tibial internal rotation during knee flexion, which the referenced papers do not report. *In-vivo* knee kinematics was acquired during quasi-static weight-bearing flexion (single-legged lunge using the free leg for stability). Conversely, during *ex-vivo* studies, loaded knee flexion was obtained by applying a force on the fixed femur (to the quadriceps tendon or to the patella, when the patellar ligament was intact) and leaving the tibia free to move and rotate, or similarly, fixing the tibia and allowing femur rotation.

Results on the deep and superficial bundle of the MCL must be interpreted in the light of the following considerations. First, typical of the MCL is the critical identification of its tibial origin, due to the lack of clear-cut bony prominences (Liu et al., 2010). Second, most of the previous studies describe the length of the ligament by a broken line wrapping around the most prominent edge of the tibial plateau (Van de Velde, et al., 2007). Conversely, in the present work, consistently with the proposed kinematic knee model, the length was computed as a Euclidean distance. The MCLsup length increased or decreased depending on the selected Monte Carlo pairs. This behaviour matches the results obtained *in-vivo* by Van de Velde et al. (2007) who showed that, during knee flexion, the central, anterior, and posterior bundles exhibited a nearly constant, increasing and decreasing length, respectively. The average tendency of the MCLdeep length to increase during knee flexion, accompanied by a maximal 10% shortening and 20% lengthening of certain Monte Carlo pairs, is in partial in agreement with the results reported by Van de Velde et al. (2007). These authors showed that the central bundle does not change significantly its length, while the anterior bundle, on average, has a 10% lengthening and the posterior bundle a 10% shortening.

The expertise and experience of the operators involved in this study is supported by the low intra- and inter-operator variability in the identification of the ligament attachment points, which was less than 2 mm. The Monte Carlo simulation included a wide range of origin-insertion pairs and confirmed the important effects of ligament attachment sites location on ligament length variation. This has been shown previously, however, using fewer origin-insertion points. Feeley et al. (2009) considered, for the MCLsup, five points on the femur and four on the tibia, and others (Grood, et al., 1989; Schutzer, Christen, & Jakob, 1989) included less than ten combinations for the ACL and PCL.

Limitations of this study were that the ligaments involved belonged to elderly individuals, with biased mechanical properties, and that an *ex-vivo* experimental model was used. It may be assumed however, that *in-vivo* joint kinematics of healthy subjects, under whatever external loading, is not different from the kinematics obtained in this study to an extent that would cause a significant change in ligament length behaviour. In both cases, in fact, stability or coaptation of the joint should be guaranteed, although in a different manner, by a synergic activity of both passive and active intra- and periarticular structures. This hypothesis is supported by recent findings that, during a highly dynamic motion such as the landing phase of a jump, showed that the ACL length decreases during flexion by an amount similar to that obtained in the present study (Taylor, et al., 2011).

In conclusion, the results of this study indicate that, in the framework of the MBO approach, a kinematic model of the knee based on joint constraints should consider the length of ACL, LCL and MCLdeep variable as a function of knee flexion. Given the dependence of PCL and MCLsup length variation from the selected attachment sites, these ligaments could be considered isometric. The efficacy of this kinematic model, as opposed to those already implemented, must be evaluated in terms of consequences on the estimate of joint kinetics, particularly when the inertial effects of soft tissue masses are involved.

4.3 HIGH RESOLUTION APPROACH: DISCUSSION

The purpose of the High Resolution Approach section was to contribute improving the accuracy of inverse dynamics methods for the estimation of joint forces. The latter, in fact, are considered to be crucial in the assessment of injury-related biomechanical variables in sports contexts. The evaluation of knee joint forces, in particular, is very important during sprint running, being the knee the most frequent site of injuries in this sports discipline.

In the framework of the multi-body optimization approach and to further improve the quality of constraint-based kinematic models of the knee, subject-specific, non-rigid constraints to the distances between the origin and insertion landmarks of the four major knee ligaments were defined. The inaccuracies associated with the estimation of the ligament attachment locations were also considered.

Results indicated that, in a kinematic model of the knee based on joint constraints, the length of the ACL, LCL and MCLdeep should be considered as variable during knee flexion, while the length of the PCL and of the MCLsup proved to be highly dependent from the selected attachment sites. These ligaments could be, therefore, considered isometric during knee flexion.

The efficacy of this kinematic model, as opposed to those already implemented, must be evaluated in terms of consequences on the estimate of joint kinetics, particularly when the inertial effects of soft tissue masses are involved.

In addition, considering that the MBO technique aims at providing an optimal estimation of the 3D position of the entire chain of bones in analysis, the reconstruction of the movement of the soft tissue masses relative to the underlying bone may also be attempted. It is worthwhile to underline, in fact, that such movement is not an artefact in itself, but that it produces artefacts when it is used to estimate the motion of the underlying bones. From a different perspective, it can give unbiased information regarding the motion of some portions of soft tissues and may be fed into a biodynamic model of the human body. The kinetic analysis of this model would incorporate the relevant inertial effects that are known to significantly affect the forces involved in motor acts characterized by high accelerations, typical of sports applications. Preliminary tests showed that using a large number of floating markers positioned on *ad-hoc* portions of the soft tissue masses, thus exploiting redundancy of the measured information, appears to be promising.

CHAPTER 5

CONCLUSIONS

Sports biomechanics describes the human movement from a performance enhancement and an injury reduction perspective. In this respect, the purpose of sports scientists is to support coaches and physicians with reliable and usable information about athletes' technique.

The lack of methods and protocols allowing for in-field athlete evaluation as well as for accurate joint forces estimate represents, to date, the main limitation to this purpose.

The results and the considerations carried out in the present thesis aim at providing a contribution towards the development of the above mentioned methods.

Results obtained from the Low Resolution Approach indicate that, due to their portability and inexpensiveness, inertial measurement units proved to be a valid alternative to traditional laboratory-based instrumentations, such as motion analysis systems and force platforms. Their usability, moreover, contributes to broaden the scope of application, allowing even coaches and runners to access information regarding the influence of running mechanics on performance and injuries.

Using acceleration and angular velocity measurements provided by the inertial units, the following quantities were estimated: trunk inclination and angular velocity, instantaneous horizontal velocity and displacement of a point approximating the centre of mass, and stride and support phase durations.

The extraction of movement-related information from the measured signals, however, is still demanding due to the movement of soft tissues relative to the underlying skeleton and to the unstable bias characterising the sensors signals. The latter proved to be relevant when numerical integration of the measured signals is necessary. To mitigate these sources of error, particularly relevant when dealing with sports motor acts, good-practice rules are recommended for the benefit of signal processing. In particular, careful attention has to be paid to the location and method of fixation of the sensor for limiting the motion of the soft tissue masses. The use of memory foam materials and elastic belts appears to be effective. To limit the errors yielded by the unstable bias of the signal, two approaches can be used: reducing the integration interval, so that to limit the amount of time in which errors accumulate, and exploring boundary conditions which can be used to cyclically correct the drift errors, particularly, in non steady-state movements.

In the framework of the multi-body optimization, High Resolution Approach results indicate that, in a kinematic model of the knee based on joint constraints, the length of the anterior cruciate, the lateral collateral and the deep bundle of the medial collateral ligaments should be considered as variable during knee flexion. The length of the posterior cruciate and of the superficial bundle of the medial collateral ligaments was found to be highly dependent from the selected attachment sites, and could be, therefore, considered isometric during knee flexion.

These results represent a first contribution to the definition of methods aiming at improving the accuracy of inverse dynamic estimates. The efficacy of a kinematic knee model, as obtained by implementing constraints that are anatomy-based, non-rigid and subject-specific must be evaluated, indeed, in terms of consequences on the estimate of joint kinetics, particularly when the inertial effects of soft tissue masses are involved.

The investment of resources aimed at improving the MBO approach, however, is justified also by the following considerations: as the MBO techniques aims at providing an optimal estimation of the 3D position of an entire chain of bones (for instance pelvis, femur and tibia), the reconstruction of the movement of the soft tissue masses relative to the underlying bone may be attempted. Such movement may be then used to develop a biodynamic model of the human body, which would incorporate the inertial effects known to significantly affect the forces involved in motor acts characterized by high accelerations, typical of sports applications. In this respect, using a large number of floating markers positioned on *ad-hoc* areas of the soft tissue masses, thus exploiting redundancy of measured information appears to be promising.

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